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CHANGES IN GAIT OVER TIME IN RESPONSE TO EXERCISE

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Statement of Originality

I hereby certify that all the work described in this thesis is the original work of the author. Any published (or unpublished) ideas, techniques, or both from the work of others are fully acknowledged by the standard referencing practices.

Camila Oliveira

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Ethical Disclaimer

Ethical approval for the studies mentioned in this thesis has been granted by the Ethics Committee of Faculty of Sport, University of Porto (Process number: CEFADE 10/2014).

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"Gratitude is not only the greatest of virtues, but the parent of all the others".
(Cicero)

List of publications

This Doctoral Thesis is based on the following scientific papers, which are referred in the text by their Arabic numbers:

1. Oliveira, C.F., Vieira E., Machado, L.J., Sousa, F., Vilas-Boas, J.P. Kinematics changes during fast walking in old and young adults. Manuscript submitted for publication.
2. Oliveira C.F., Soares D.P., Bertani M.C., Rodrigues Machado L.J., Vilas-Boas J.P. Effects of Fast-Walking on Muscle Activation in Young Adults and Elderly Persons. *Journal of Novel Physiotherapy and Rehabilitation*. 2017; 1: 012-019.
3. Oliveira, C.F., Vieira, E., Sousa F., Machado, L.J., Vilas-Boas, J.P. Age-related changes in kinematic and muscle activation variability during fast-walking. Manuscript submitted for publication.
4. Oliveira, C.F., Soares D.P., Sousa, F., Machado, L.J., Vilas-Boas, J.P. The influence of fast-walking on the variability of slip and trip-related kinematics. Manuscript submitted for publication.
5. Oliveira, C.F., Vieira, E., Machado, L.J., Sousa, F., Vilas-Boas, J.P. Older adults make use of different strategies to control toe clearance during fast-walking. Manuscript submitted for publication.

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Resumo

Os efeitos da caminhada rápida nos parâmetros da marcha parecem destacar potenciais ameaças durante a execução desta atividade. Assim, o objetivo geral deste estudo foi investigar as alterações da marcha durante um exercício esporadicamente executado pelos idosos, a caminhada rápida. Durante esta atividade, dados cinemáticos e eletromiográficos foram avaliados. Os idosos apresentaram alterações nas articulações do quadril e do tornozelo em instantes discretos do ciclo de marcha sugerindo uma mudança na estratégia de produção de energia para impulsionar o corpo. Em geral, as alterações cinemáticas encontradas foram acompanhadas por mudanças no padrão de ativação muscular do membro inferior. Durante a tarefa, a atividade do músculo tibial anterior aumentou durante a fase de absorção e reduziu durante a sua fase propulsiva. Ao longo do protocolo, os índices de co-activação mudaram significativamente. Padrões de ativação muscular do tibial anterior e do gastrocnémio medial, reduziram e as regiões de maior variabilidade dentro do ciclo de marcha mudaram. Estas alterações neuromusculares foram semelhantes às encontradas na cinemática angular. Além disso, as reduções observadas na atividade muscular, níveis de coativação e variabilidade no padrão de ativação muscular, sugerem uma tentativa de redução do consumo metabólico durante o exercício. A presente pesquisa contribui para o aumento do conhecimento nesta área, fornecendo informações adicionais sobre a presença de variabilidade dentro do ciclo de marcha. Adicionalmente, encontramos mudanças nessas regiões de maior variabilidade dentro do ciclo de marcha durante a caminhada rápida. Estas alterações sugerem que adaptações a perturbações externas são dependente da fase do ciclo de marcha e podem ser consideradas como uma tentativa para manter o padrão motor dentro dos limites de segurança. Além disso, dada a associação encontrada entre a cinemática angular e o controle do movimento da trajetória vertical do pé e da velocidade horizontal do calcanhar, estratégias para prevenir quedas por tropeçar ou escorregar devem priorizar a estabilidade articular do tornozelo e do joelho. No entanto, apesar de melhorar o controle do movimento em eventos críticos do ciclo de marcha, os idosos parecem ter aumentado o risco de queda enquanto caminhavam em ritmos mais acelerados.

PALAVRAS-CHAVE: CINEMÁTICA, FADIGA, CAMINHADA, TROPEÇO, QUEDAS EM IDOSOS

Abstract

Understanding the ongoing effects of fast-walking activity on gait parameters may highlight potential threats that can arise while walking at a faster pace. Therefore, our general purpose was to investigate the alterations to gait during an activity that can be sporadically performed by the elderly such as fast-walking. During the activity, kinematics and electromyography data have been assessed. Old adults showed alterations at the hip and ankle at discrete moments of the gait cycle suggestive of a change in the power generation strategy for propelling the body forward. Moreover, the alterations found on gait kinematics due to fast-walking were followed by changes in muscular activation pattern of the lower limb. During the task, the activity of the tibialis anterior increased during the absorptive phase and reduced during the propulsive phase. Throughout the protocol, coactivation indexes significantly changed over time. Measures of interstride variability of muscle activation for tibialis anterior and gastrocnemius medialis reduced, and regions of intra-stride variability changed during fast-walking. These neuromuscular alterations were similar to those of the angular kinematics. The reductions observed in muscular activity, levels of coactivation and variability in muscular activation pattern could be interpreted as an attempt to reduce metabolic consumption during the activity. The research presented here contributes to the body of literature by providing additional information regarding the presence of intra-cycle variability. Additionally, we have also found changes in these regions of higher variability within the gait cycle during fast-walking. These changes suggest that adaptations to a perturbation are phase-dependent and can be considered as an attempt to maintain the motor pattern within safety boundaries. Moreover, given the association found between the angular kinematics and the movement control of the toe vertical trajectory and heel horizontal velocity, strategies to prevent falls by tripping or slipping should address ankle joint kinematics, also as the knee joint kinematics. Despite improving movement control at critical events of the gait cycle, the elderly may have increased the risk of fall by tripping or slipping while walking at faster pace.

KEY WORDS: KINEMATICS, FATIGUE, WALKING, TRIPPING, FALLS IN ELDERLY

List of Abbreviations and symbols

ADL	Activity of daily living
ANOVA	Analysis of variance
BF	Biceps femoris
CAPES	Coordination for the Improvement of Higher Education Personnel
CoI	Coactivation index
COM	Center of mass
EMG	Electromyography
GM	Gastrocnemius medialis
HR	Heart rate
HCV	Heel contact velocity
HHV	Heel horizontal velocity
IBGE	Brazilian Institute of Geography and Statistics
INE	National Statistical Institute of Portugal
IQR	Interquartile range
MTC	Minimum toe clearance
n	Number of subjects
OA	Old adults
PC1, PC2, PCn	First, second, n_{th} principal component (PCA)
PCA	Principal component analysis
ROM	Range of motion
RMS	Root mean square
SD	Standard deviation
SPSS	Statistical package for the social sciences
TA	Tibialis anterior
TVD	Toe vertical displacement
U	Eigenvectors of the covariance matrix (PCA)
VM	Vastus medialis
WHO	World Health Organization
YA	Young adults
β	Coefficient of regression
Δ	Deviance

λ	Eigenvalues (PCA)
3D	Three-dimensional

Currently, society is facing a marked ageing of the world population. Approximately 14% of the European population are over 65 years and it is expected that these numbers will double by 2050 (WHO, 2011). In Portugal, between 2001 and 2011, the percentage of the elderly population increased from 16.6% to 19% (INE, 2011). In Brazil, currently, about 10% of the population are elderly, i.e., approximately 20 million people (IBGE, 2010). With more than a third of elderly experiencing at least one fall per year worldwide, the consequences of falls represent a major problem for the health systems of many countries with huge financial costs associated (WHO, 2011).

Falls are a result of a complex interaction between extrinsic (environment) and intrinsic (related to the individual) factors. In a systematic review regarding the association of biomechanical aspects of gait in elderly fallers and non-fallers, Kirkwood et al. (2006) observed that 55% of falls were related to gait alterations, 32% to balance alterations, and the remaining to extrinsic factors. Among the extrinsic factors found were: inappropriate shoes, uneven surfaces, poor lighting, etc. The intrinsic factors included poor health, lack of static and dynamic postural control (i.e. remain standing and walking), insufficient muscle strength and power of the lower limb muscles, visual difficulties, history of falls, and fatigue (Maki, 1997; WHO, 2011). There are currently over than 400 known risk factors for falls (Masud & Morris, 2001). Besides the extrinsic (environmental) and intrinsic (individual) factors, aspects related to the task are among those that can bring an individual to fall. Factors such as speed, task complexity or fatigue during the task are considered to increase the probability of a fall incident (Callisaya et al., 2011; Tinetti & Speechley, 1989).

Healthy, active seniors are more likely to fall in outdoor activities where walking faster can be sporadically performed by this population (Li et al., 2006). Research supports the premise that age-related differences on gait observed in elderly are primarily due to reduced muscle strength and lower limb joint range of motion (ROM) (Kang & Dingwell, 2008), which in turn, is attributable to physiological and neuromuscular changes displayed by older adults (Faulkner et al., 2007; Prince et al., 1997). Therefore, the extent of these changes may reduce the capacity of this population to adapt their gait pattern when demanded. Given that walking biomechanics are known to be influenced by speed, some studies have investigated the effects of the walking speed as well as the underlying factors that may drive these alterations. Significant associations have been found between gait speed and joint kinematics (Hanlon & Anderson, 2006; Monaco et al., 2009), joint kinetics (Burnfield et al., 2000; Chung & Wang, 2010; Silder et al., 2008), muscle activity (Schmitz et al., 2009) and gait stability (England & Granata, 2007; Kang & Dingwell, 2008). Irrespective of the factors underlying the alterations caused by walking at a faster pace, some authors have associated fast-walking with increased risk of falls (Callisaya et al., 2011; Faulkner et al., 2009; Nagano et al., 2014; Pavol et al., 1999).

A successful locomotion needs the integration of different physiological systems. According to England and Granata (2007) fast-walking velocity may influence the dynamic stability by a combination of several mechanisms, as the ability to control movement could be disrupted by the effects of fast-walking over gait kinematics and other clinical correlates of stable walking. When walking, the ability to adjust the speed requires different levels of muscular activities for appropriate adaptations to changes in the task demands. Nevertheless, increasing walking speed is associated with an increase in muscle stress, particularly of the plantar-flexors and dorsi-flexors (Neptune, 2004). Indeed, the ability of the ankle plantar flexors to produce force as walking speed increased was greatly impaired (Neptune et al., 2008), which could potentially compromise the trajectory of the swing foot, increasing risks of tripping and slipping (Lockhart & Kim, 2006; Winter, 1992). In consonance with this, Nagano et al. (2014) demonstrated that after a short period of fast-walking, older adults were more susceptible to falls by tripping, the authors attributed this to fatigue. In addition, older adults increased coactivation at the knee and ankle during mid-stance when walking faster than the preferred speed. Whilst coactivation is believed to be used to stiffen the joint and enhance stability (Hortobágyi & DeVita, 2006), a potential side-effect would be higher metabolic costs (Mian et al., 2006). Furthermore, increased joint stiffness decreases the capacity to produce force during toe off

(Watelain et al., 2000), as well as increases horizontal heel velocity at foot landing (Lockhart & Kim, 2006). Therefore, the broad effects of increased coactivation in older adults can increase the risk of falling by tripping or slipping.

The great variety of fall-risk's factors reflect the diversity of health determinants that directly or indirectly affect the individual's well-being. Increased speed and walking duration may magnify the limitations of old adult's gait characteristics, changes may occur over time, and such assessments may highlight other factors that may be associated with risk of falls. Studies involving analyses of biomechanical parameters and muscle activity in the elderly during an activity that can be a sporadic practice among this population, such as fast walking, are crucial for understanding the strategies that may be involved in falls' prevention. Thus, there is a growing need to prevent falls in the elderly; to present relevant information regarding the abnormal gait patterns induced by fast-walking in the elderly; to determine mechanisms of falls allowing the development of a feedback mechanism in order to prevent the occurrence of falls and reduce such alarming numbers of morbidity associated with this part of the population; to give a differential treatment of the information obtained, considering data under different levels as well as the possible interactions between them; and to provide further information to public health systems to allow the planning of strategies addressed to the current reality, taking into account the context of the increase in life expectancy in most countries.

Objectives

Based on these assumptions, we observed a lack of information regarding the emerging changes to gait in an ecological context that may determine the predictive behavior of falls in the elderly population. The overall aim of this thesis is to evaluate the effects of fast-walking on gait, with a special focus on evaluating parameters previously linked to risk of falls in older adults. The specific aims were as follows: (1) to describe alterations on gait kinematics during fast-walking in old and young adults; (2) to investigate the influence of fast-walking on muscle activity of lower limb muscles of young and older adults; (3) to determine how the variability of gait and muscle activity change during prolonged time walking at faster speed; (4) to determine whether the variability of EMG and kinematic patterns during fast-walking were reflected in the variability of kinematics associated with risk of slipping and tripping; (5) to verify alterations on slip and trip-related parameters during the activity; (6) to determine the influence of the swing and stance lower limb joints on altering the swing toe trajectory.

The work presented here aims to determine the biomechanical gait changes induced by fast-walking in elderly, with the experimental accomplishments presented in Chapters 3 to 7. In Chapter 2 we provide an overlook of the scientific literature pertaining to the topic. Then, we describe in Chapter 3 kinematic alterations over time in both age groups. Chapter 4 explores neuromuscular adaptations during fast-walking in old and young adults. Chapter 5 explores the variability of muscle and kinematic pattern during the task. In chapter 6 we sought to investigate whether the variability of trip and slip-related signals were associated with those from muscle and kinematic pattern described in chapter 5. Chapter 7 reveals the risk of falls by tripping along the fast-walking protocol. Chapter 8 includes a general discussion in the context of the findings from the experimental studies. Chapter 9 presents the main conclusions and offers recommendations for future research. Finally, the bibliography references are presented in Chapter 10. An overview of study designs is presented in Table 1.

Table 1: Study design description.

Study	Design	Population	n included
I	Descriptive analysis of gait kynematics alterations over time	Healthy young and older adults	50
II	Muscle activation study	Healthy young and older adults	23
III	Variability of EMG and kinematic patterns	Healthy young and older adults	23
IV	Gait variability influence on slip and trip-related Association	Healthy older adults	08
V	Multilevel Regression Analysis	Healthy young and older adults	51

n = Number of subjects.

Study delimitation

The study was delimited to healthy old and young adults above 65 and 21 years old, respectively. All participants were free of any orthopaedical, neurological, visual, vestibular or cardiovascular conditions that would prevent the subject from performing all the proposed activities. Old participants were recruited from the local community and the young participants from among the university's students. The walking-based fatigue protocol consisted of walking at 70% of their maximum heart rate for twenty minutes or until their voluntary exhaustion. The flow of this protocol is illustrated in Figure 1. Further explanations regarding data acquisition and analysis are described in the experimental papers throughout the thesis.

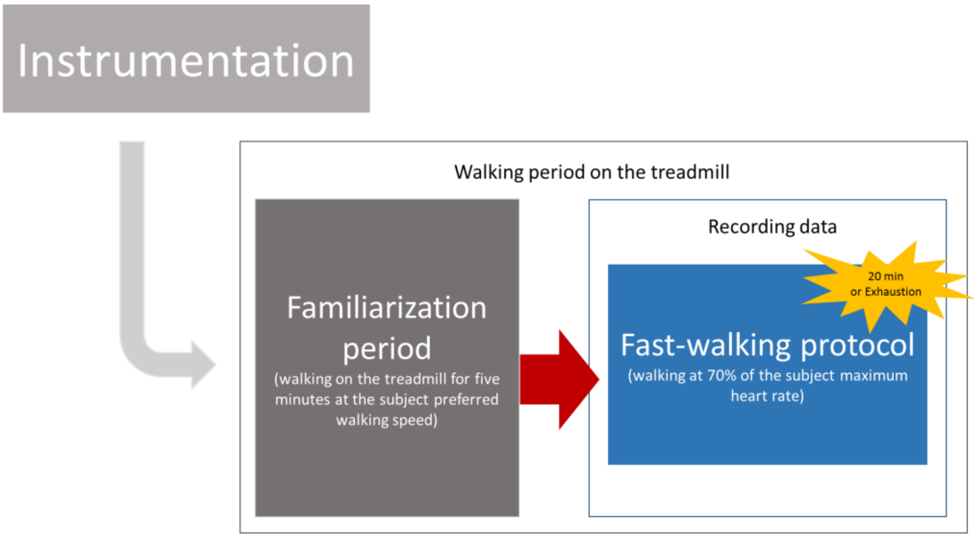

Figure 1: Workflow of the fast-walking protocol

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Chapter 2 Literature Review

This chapter reviews the scientific literature pertaining the topic being investigated in four major sections: incidence and aetiology of risk of falls in the elderly population; a comprehensive review of fatigue in older adults; and a summary about fatigue and increased speed effects on gait parameters.

2.1 Falls in elderly population

Ageing of the population and its arising implications led the construction of a new epidemiological profile, characterized by a higher prevalence of chronic diseases, bringing new challenges in health care (Williams & Manfredi, 2004). According to WHO (2011) more than a third of elderly experienced at least one fall per year. A study carried out in Catalonia (Spain) revealed that 17.9% of people over 65 have experienced at least one fall in the 12 months preceding the interview and the falls frequency increases with age (Séculi et al., 2004). In Portugal, after direct consultation with the National Statistical Institute (INE) and the Central Administration of the Health System (ACSS), no official data were found regarding the prevalence of falls in the elderly. In Brazil, the falls and their consequences in the elderly have assumed epidemic dimension, with huge financial costs associated with (IBGE, 2010).

The combination between high incidence with a high susceptibility to injury makes falls in the elderly population particularly dangerous, because of the high prevalence of clinical diseases (e.g. osteoporosis) and age-related physiological changes (e.g. slower reflexes) in this population (Rubenstein, 2006). In addition, the rate of falls increases with age, and the fear of falling down-regulates activity, which in turn, further contributes to subsequent falls (Tinetti & Powell, 1993).

Despite being the leading cause of non-fatal injuries worldwide, and therefore motivating several research studies focusing on fall prevention, there remains a lack of detailed national estimates of falls incidence, and fall-related injuries by age and gender. Data on the older Portuguese population relative to this matter is scarce, however Moniz-Pereira et al. (2013) in a prospective study reported a first known faller risk profile in the Portuguese elderly population. Their results showed that being a woman, having fear of falling and lower functional fitness levels were determinant factors for both episodic and recurrent falls.

Given the high incidence of falls and the associated morbidity, mortality and other costs, combined with the ageing of the population, the incidence of falls in the elderly and the associated costs are expected to rise unless effective preventive techniques are implemented. If preventive measures are not taken in the immediate future, the numbers of injuries caused by falls are projected to be 100% higher in the 2030 (Kannus et al., 2007).

Falls occur as a result of a complex interaction between numerous environmental hazards and individual factors. Among the extrinsic factors, we found inappropriate shoes, uneven surfaces, poor lighting, etc. The intrinsic factors included poor health, lack of static and dynamic postural control (i.e. remain standing and walking), insufficient muscle strength and power of the lower limb muscles, visual difficulties, and history of falls (WHO, 2008). The presence of environmental hazards creates the opportunity for a fall, particularly for individuals already impaired by a combination of intrinsic factors (Rubenstein, 2006). Gait or balance deficits emerged as the most consistent independent predictors from a systematic review regarding falls risk factors (Ganz et al., 2007). Ageing brings about profound changes in human locomotion. Older adults seem to adopt a more conservative walking pattern characterised by a slower walking velocity, shorter steps, greater base of support and prolonged double support phase (Lord & Dayhew, 2001; Maki, 1997). It is generally assumed that these changes lead to an increased stability during walking (Lord & Dayhew, 2001). One critical factor closely related with the task of walking itself is fatigue, which may discourage physical activity and further compromise safe progression. Nevertheless, several studies have identified fatigue as an important risk of falling, due to the deleterious influence in several aspects of gait modified by fatigue (Helbostad et al., 2007; Parijat & Lockhart, 2008b).

2.2 Fatigue in older adults

Given the ageing of our population and the improvements in medicine that have extended the life expectancy of human beings over the last century, the investigation about whether and how skeletal muscles of the elderly resist fatigue has become an important area of research. In an intact organism, several factors can influence the development of muscle fatigue. Because the ability to maintain force is a critical aspect of neuromuscular function, understanding fatigue is essential to understand the biology of senescence (Kent-Braun, 2009). Some changes in the motor system related to age have an impact on the resistance to muscle fatigue. Examples of changes are:

- Loss and atrophy of muscle fibers, with a great decrease in the number and size of type II muscle fibers (Campbell et al., 1973);

- Progressive decline in the number of motoneurons with subsequent re-innervation and expansion of the innervation area of the surviving motoneurons (Manini et al., 2013);

- Substantial loss of upper motoneurons in the central nervous system (Eisen et al., 1991).

The functional consequences of these changes include substantial loss of strength and decreased contractile characteristics in most muscles assessed. Further studies suggest that type I muscle fibres contribute proportionally more to the strength production in elderly (≥70 years) than in young people (Roos et al., 1997). However, although it seems reasonable to suggest that age-related changes in muscle morphology and motor unit remodelling, as well as associated loss of strength and contractile properties, may improve resistance to neuromuscular fatigue in elderly, the results highlighted and discussed in a literature review performed by Allman and Rice (2002) suggest that this generalization cannot be made. The lack of agreement between the studies is due, in part, to the differences by which fatigue was induced or assessed. These inconsistencies may result in the quite varied methodologies used to quantify fatigue, as well as the different muscle groups that have been tested (i.e., elbow flexors, knee extensors, thumb adductor, ankle dorsiflexors), the exercise modality selected (i.e., isometric, isokinetic, dynamic, with electrical stimulation fatigue), the particularities of the study populations, and the relatively small number of individuals studied (Katsiaras et al., 2005).

The variability of fatigue susceptibility in the elderly relies heavily on the task used to induce fatigue. The fatigability in the elderly depends on the performance of different types of exercises (voluntary vs. electric, isometric vs.

dynamic, sustained versus intermittent, maximal vs submaximal) and yet, on the performance of the same exercise by muscles with different contractile properties (Bigland-Ritchie et al., 1995). According to Chan et al. (2000), muscle fatigue can result from the failure of any component along the motor pathway, from the volitional effort to the contractile ability of the muscle. This approach resulted in the distinction between central and peripheral fatigue, or central nervous system and muscle fatigue. Thus, the concept of fatigue begins to consider its multifactorial nature involving issues related to the depletion of energy systems, the accumulation of catabolism products, the involvement of the nervous system and the failure of the contractile mechanism of skeletal muscle fibre. The central fatigue was also considered as the failure to activate all motor units at optimal discharge rates and it can be assessed by comparing the reduction in force during a maximal voluntary contraction, which requires full functioning of the entire muscle force production pathway, to the decreased force induced by electrical stimulation, which evaluates changes distal to the stimulation point (Kent-Braun, 2009). For clinical use, the central fatigue is defined as the difficulty of initiation, or the ability to maintain volunteer activities (Chaudhuri & Behan, 2000). Finally, the ability to resist fatigue, sometimes expressed as 'muscle endurance', was defined as the time to failure to maintain target tension (Hicks et al., 2001).

In contrast to the concept of muscle fatigue suggested by Enoka and Stuart (1992), as an acute impairment of performance that includes both an increase in the perceived effort necessary to exert a desired force and an eventual inability to produce this force, a new review carried out after 15 years by Barry and Enoka (2007) defines muscle fatigue as an exercise-induced reduction in the ability of the muscle to generate force or power, even if the task can still be sustained. Therefore, fatigue often begins soon after the onset of sustained activity, even though an individual can continue performing the task. Although the impairments that contribute to fatigue will eventually limit the ability of the individual to continue that task, fatigue and task failure should be distinguished (Bigland-Ritchie et al., 1995).

Considering aspects of different systems, some studies observed that in the elderly population, when compared to young adults, presented greater resistance to fatigue. By incorporating information from multiple systems, the fatigue resistance observed by the elderly seems to lie on a neuro-energetic basis centered on the largest metabolic economy in the neuromuscular system (Lanza et al., 2005). Regarding fatigue in the elderly, age-related changes that occur within the neuromuscular system may result in some sites that are more prone to fatigue. The changes may increase or decrease their susceptibility to failure under specific task conditions. The effect of age on the various central and peripheral sites more prone to fatigue is discussed considering their relative contributions during the different fatigue-induced tasks, without neglecting the impact of possible confounding effects on fatigability related to subjects' habits, physical activity status and gender (Allman & Rice, 2002). The assumption obtained after the results of more recent studies, presented in a review performed by Kent-Braun (2009), indicate that energy-producing pathways in skeletal muscle may combine with changes in motor unit behaviour and in contract properties of muscle with the objective to provide a unique environment to resist the muscle fatigue in healthy men and women over the age of 65.

2.3 Fatigue effects on gait and its repercussion on risk of falls

Fatigue can have a widespread impact on biological functions, altering the capacity of most systems to operate at the desired level (Enoka & Stuart, 1992; Gandevia, 1998). The effects of fatigue on gait in elderly people have been a crucial focus of some recent studies.

The impact of fatigue is not restricted to a decline in force producing capacity of the system, but also it leads to an inability to adequately control specific movement dynamics to produce a controlled movement (Cortes et al., 2014). A certain level of variability associated with many biomechanical measures contains structure in the form of long range correlations. This means that a repeated movement at any point in the time series is related to or dependent upon previous cycles. This scaling behaviour in human gait is of interest because it indicates the overall adaptability of the motor system. Increased variability may bring the dynamic state to their limits of stability, where error corrections are less effective. In this context, increased variability has been found to be associated with fall risk (Kressig et al., 2008). In an exemple, Helbostad et al. (2007) reported that a repeated sit-to-stand task affected gait control in older person in terms of an increased variability in step width and length.

Stable walking kinematics requires generating appropriate motor patterns (Lee & Kerrigan, 1999). The movement pattern presumably reflects that of the muscles. However, Granacher et al. (2010) reported that ankle fatigue resulted in a decrease in functional reflex activity of tibialis anterior, and increases in coactivity and mean angular velocity in the ankle joint complex. It has been suggested that increased coactivation reduces the capacity to produce force in a short period of time, which could lead to lower capacity to react to motor disturbances (Bellew, 2002). A later study demonstrated coactivation as an important risk factor for falling (Pereira & Gonçalves, 2011), since it negatively affects torque production and increases energy expenditure during gait in elderly. The authors suggested that these results may impair an individual's ability to compensate for gait perturbations. Appropriate temporal separation between agonist and antagonist activation of muscles has been observed for well-controlled voluntary movements (Frey-Law & Avin, 2013). When fatigated, however, this separation is attenuated, muscle coactivation is increased reducing knee extensor force (Pereira & Goncalves, 2011). Furthermore, higher coactivation increases joint stiffness decreasing the capacity to produce force during toe off (Watelain et al., 2000), as well as also increasing horizontal heel velocity at foot landing (Lockhart & Kim, 2006). Therefore, since loss of dorsiflexor strength at toe-off can be critical to the ability of overcome an obstacle (Nagano et al., 2011), and higher horizontal heel velocity at landing increases chances of slips (Lockhart & Kim, 2006), the broad effects of increased coactivation in older adults can increase risk of falling by tripping or slipping.

As previously mentioned, fatigue can lead to a loss of muscle strength, which may compromise obstacle clearance. The safe trajectory of the foot during swing is essential to the successful obstacle avoidance. The swing toe describes its forward trajectory reaching its lowest vertical point, represented by the minimum toe clearance (MTC). This biomechanical event occurs at the highest forward velocity of the swing foot, representing a highly-controlled movement to maintain the clearance around a safety value and prevent tripping (Winter, 1992).

Previous work has explored fatigue influence on the MTC and reported significant reductions in toe clearance, which in turn increases risk of falls by tripping. Along with tripping, slipping have also been shown to contribute to high rates of falls. The knee joint musculature is considered important in producing large flexion and extension moments when recovering from a slip. Fatigue of the knee extensors is associated with decreases in stabilization time (Parijat & Lockhart, 2008b).

As successful locomotion needs the integration of different physiological systems, therefore many factors have influence on gait performance. The inevitable consequences of fatigue can alter neuromuscular processes both centrally and peripherally. Since walking requires contracting muscles to move an imposed load through an adequate joint ROM for successful task completion, fatigue can affect the movement performance for a given individual. Previous studies indicated that slower walking speed in elderly subjects may be associated with the physiological or neuromuscular limitations in older adults (Anderson & Madigan, 2014; McGibbon, 2003). Regarding fatigue effects on joint angular kinematics, previous studies reported reduction in ankle joint ROM, greater knee flexion at heel contact, less knee extension during terminal stance, and less dorsiflexion at heel contact (Cheng & Rice, 2012; Parijat & Lockhart, 2008b). The reduction in the ROM of one distal joint, which has been reported as a consequence of fatigue, can lead the adjacent joints to adapt to a greater extent, producing different kinematic combination. The resulting coordination pattern of the whole limb, therefore, might have to adapt in a non-conventional way to accomplish the motor task required. However, the studies have been limited to a description of the fatigue effects on singular joints, without considering the intersegmental interactions.

In summary, fatigue can affect sensorimotor function and walking performance in many ways. Ultimately, fatigue has been shown to increase risk of falls by increasing gait variability, reducing minimum toe clearance, and increasing slip risk. Furthermore, when exposed to fatigue, the sensorimotor system seems to adjust motor planning and execution at the expense of the angular kinematics pattern to ensure the success of a motor task (Cheng & Rice, 2012; Parijat & Lockhart, 2008a). However, most of the known effects of fatigue on gait parameters were assessed after using protocols that are not commonly reproducible in daily life. Recently, walking-based fatigue has been used to induce fatigue, although most of the studies have been limited to preferred walking speed. The only known exception regarding the association between fatigue and risk of falls is the study of Nagano et al. (2014), which have demonstrated that after a short period of fast-walking, older adults were more susceptible to falls by tripping.

2.4 Effects of fast speed on gait

In general, healthy older adults are more susceptible to fall in outdoors activities (Li et al., 2006), where walking at a faster pace can be a sporadic practice among this population. Previous experiments have demonstrated that walking at a faster speed significantly affect gait biomechanics. In a systematic review by Figueiredo et al (2011), the review showed the influence of speed in biomechanical parameters that characterize the gait action. Significant associations have been found between gait speed and joint kinematics (Hanlon & Anderson, 2006; Monaco et al., 2009), joint kinetics (Burnfield et al., 2000; Chung & Wang, 2010; Silder et al., 2008), muscle activity (Schmitz et al., 2009) and gait stability (England & Granata, 2007; Kang & Dingwell, 2008).

Previous studies have reported that a higher walking speed caused a greater joint motion in hip and knee joints (Hanlon & Anderson, 2006; Lelas et al., 2003; Monaco et al., 2009). In Hanlon and Anderson (2006), it is brought to our attention that the degree to which speed affected each angle depends not only on the angle itself but also on the time within the gait cycle. This is important for addressing the underlying reasons that drive these changes within the gait cycle. Some of these alterations, as the increased peak knee flexion during pre-swing, have been attributable to the need for greater shock absorption at higher speeds (Lelas et al., 2003). In addition, due to the increasing speed, older adults could exhibit a significant increase of the kinetics at the proximal joints (Burnfield et al., 2000). Previous studies have reported that older adults exhibit a proximal joint power strategy to propel the body forward (DeVita & Hortobagyi, 2000; Neptune et al., 2004). This distal to proximal shift in power production seems to be attributable to the plantar-flexor weakness (Silder et al., 2008). Assessment of joint kinetics provides a better understanding of normal motor patterns. Thus, the estimation of the relative contribution of each joint to the total energy generated and absorbed during gait at fast-walking is a useful approach to understand the extent, and degree of compensation across joints and to address potential risks due to fast-walking.

Furthermore, when walking at a faster speed, older adults increase gait and neuromuscular variability (Almarwani et al., 2016; Kang & Dingwell, 2008; Raffalt et al., 2017), and greater joint kinematic variability has been consistently associated with risk of falls (Kobayashi et al., 2014; Mills et al., 2008). In addition, a significant association have been found between gait speed and gait stability (England & Granata, 2007; Kang & Dingwell, 2008). According to England and Granata (2007) fast-walking velocity may influence the dynamic stability by a combination of several mechanisms, as the ability to control movement could be disrupted by the effects of fast-walking over gait kinematics and other clinical correlates of stable walking.

When walking, the ability to adjust the speed requires different levels of muscular activity for appropriate adaptations to changes in the task demands. In general, muscle activity tends to increase at faster walking speeds, resulting in a larger muscular force output (Neptune et al., 2008). According to Schmitz et al. (2009), the changes in muscle activity observed when walking at a faster pace enhances the distal-to-proximal shift in power production exhibited in old adults. In addition, old adults increased coactivation at the knee and ankle during mid-stance. Whilst coactivation is believed to be used to stiffen the joint and enhance stability (Hortobágyi & DeVita, 2006), a potential side-effect would be at a higher metabolic cost (Mian et al., 2006). Nevertheless, increasing walking speed is associated with an increase in muscle stress, particularly of the plantar-flexors and dorsi-flexors (Neptune, 2004). Indeed, the ability of the ankle plantar flexors to produce force as walking speed increased was greatly impaired (Neptune et al., 2008), which could potentially compromise the trajectory of the swing foot, increasing risks of tripping (Winter, 1992). In consonance with this, as mentioned earlier in this chapter, Nagano et al. (2014) demonstrated that after a short period of fast-walking, older adults were more susceptible to falls by tripping, to which the authors attributed to fatigue.

Therefore, increments in gait speed have been shown to have a significant effect on gait in older adults. However, most of these studies are limited to describe the changes that occur when walking speed is modified. Understanding the effects of fast-walking activity over time on gait parameters may highlight potential threats that may arise while walking at a faster pace.

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Chapter 3

Kinematics changes during fast walking in old and young adults

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Abstract

Walking biomechanics is known to be influenced by speed. However, most of the research examining the effects of walking speed and gait characteristics has been conducted after the walking activity, neglecting the ongoing changes that may occur during the task. Thus, 27 young adults and 23 old adults were requested to walk at 70% of their maximum heart rate for 20 minutes or until voluntary exhaustion. To determine the impact of fast-walking, differences on spatial temporal parameters and angular kinematics were analyzed during the activity. Both age groups changed their gait strategy by reducing cadence and increasing stride length while progressively increased the step width variability. For old adults, kinematic changes were more prominent at the toe-off, with reduced hip extension and increased ankle plantarflexion. Whilst the young adults significantly reduced ankle dorsiflexion at heel strike. Angular kinematic variability assessed by a higher standard deviation during heel strike and toe off had no significant differences throughout the task. In addition, the effect of the activity in the coordination of lower limb angles were more evident during the swing phase. Alterations found for the young group were compatible with that of fatigue installation. The old group seemed to change the power generation strategy for propelling the body forward. In addition, changes in the coordination of lower limb angles suggest that effects of the activity were more evident during the swing phase. The findings are important for providing foundation on which to address further analyses regarding potential risks associated to walking at a faster pace.

KEYWORDS: KINEMATICS, ELDERLY, GAIT, FALLS RISK, FAST-WALKING

3.1 Introduction

The world population is ageing. Approximately 14% of the European population are over 65 years and it is expected that this proportion will double by 2050 (WHO, 2011). With more than a third of old adults experiencing at least one fall per year, the consequences of falls represent a major problem for the health systems of many countries and represent huge costs (WHO, 2011). Falls often occur during gait and can be a result of ageing-related changes (Faulkner et al., 2007; Ganz et al., 2007).Gait changes observed in old adults are primarily due to reduced muscle strength and lower limb joint range of motion as a result physiological and neuromuscular changes (Faulkner et al., 2007; Kang & Dingwell, 2008; Prince et al., 1997). Ageing-related changes in gait may be of particular concern when walking faster than usual. Additionally, significant associations have been found between walking speed and gait stability (Kang & Dingwell, 2008), muscle activity (Oliveira et al., 2017) and energy consumption (Peterson & Martin, 2010). Furthermore, fast-walking adversely affects walking pattern, as observed by changes in spatial-temporal gait parameters (Nagano et al., 2014) and joint kinematics (Chung & Wang, 2010; Hanlon & Anderson, 2006).

Walking is the most commonly performed activity of daily living, and walking biomechanics is known to be influenced by walking speed. Some studies have evaluated the effects of the walking speed on gait and its association with risk of falling (Callisaya et al., 2011; Fan et al., 2016; Kang & Dingwell, 2008; Nagano et al., 2014). Despite most of the previous studies reporting that fast-walking increases the risk of falling (Callisaya et al., 2011; Faulkner et al., 2009; Nagano et al., 2014; Pavol et al., 1999), some authors (Fan et al., 2016) sustained that fast-walking can be safely performed as physical activity exercise by healthy old individuals.

Increased speed and walking duration may magnify the effects or limitations that may occur over time, and such assessments may highlight other factors that may be associated with falls. However, most of the research examining the effects of walking speed and gait characteristics has been conducted after the walking activity, neglecting changes during the task. Assessment of gait variability, namely fluctuations in spatial-temporal gait characteristics, may provide additional insights about motor control of gait (Hamacher et al., 2011). In addition, little is known about the fast-walking effect on kinematics over time. Thus, the present study was conducted to determine the impact of walking at a faster pace on spatial-temporal parameters and kinematics in young and old adults during the activity.

3.2 Methods

3.2.1 Subjects

The participants were 27 young (26.6 ± 6.0 years) and 23 old adults (71.0 ± 5.6 years). They all were free of orthopaedical, musculoskeletal or cardiovascular constrains that might impair normal locomotion. All participants provided informed consent using procedures approved by the Local Ethics Committee.

3.2.2 Procedures

Kinematic data were captured (100 Hz) using an 8-camera motion capture system (Qualisys, QTM). Reflective markers were placed on the pelvis (anterior-posterior and posterior-superior iliac spines, lateral aspects of thighs, knees (femoral condyles), shanks and ankle (malleoli), and on the feet (end of the 2nd toes, 5th metatarsal head, and calcanei). Data were than processed using Visual 3D (C-motion, Rockville, MD). During the procedures, old participants wore a safety harness attached to the ceiling.

3.2.3 Protocol

All participants completed a familiarization session on a treadmill (AMTI Inc., MA, USA) walking at their preferred walking speed for five minutes. After a familiarization period of 5 minutes, the treadmill speed was increased by 0.5 km/h every 30 seconds till the participants reached the speed where they would achieve 70% of their age-predicted maximal heart rate (220 minus age in years). The participants were instructed to walk at this intensity for twenty minutes or until their voluntary exhaustion. Heart rate (HR) was measured continuously during the test with a Polar HR monitor (2010 Polar Electro Oy, FI-90440 Kempele, Finland) to ensure subjects were walking at the required intensity. Then, the relative speed was kept the same for each of the participants, and constant throughout the test. Additionally, the rating of perceived exertion (RPE) scale was obtained throughout the protocol using a modified Borg 10-point scale (Borg, 1982). During the protocol, a variation of ten percent of target heart rate was allowed. The participants would be asked to stop the protocol if the exertion reached 90% of their maximum heart rate.

Lower-limb joint angles and spatial-temporal parameters, including spatial temporal parameters, were collected for 30 seconds every minute until the end of the activity. Five equally distributed stages were chosen for analysis. Gait parameters were obtained for the dominant limb. To determine the lower-limb dominance the participants were asked to kick a ball, the leg used to kick the ball were chosen as dominant. All the participants of this study were right lower-limb dominant. Visual 3D (C-motion Inc., Rockville, MD, USA) was used to calculate kinematic data. The marker trajectories were filtered using a fourth-order low-pass Butterworth filter with a 12 Hz cut-off frequency. Gait cycle was defined by consecutive heel strikes of the right foot. Kinematics of the lower-limb angles were calculated using an XYZ Cardan sequence of rotations (where X is flexion-extension; Y is ab-adduction and is Z is internal-external rotation). Hip flexion, knee flexion, and dorsiflexion were classified as a positive displacement. Whereas, hip extension, knee extension and plantarflexion were determined as a negative displacement.

The following parameters were assessed: sagittal angle of the hip, knee and ankle at heel strike and toe off, cadence (number of gait cycles per minute), stride length (distance between two successive heel strikes), stride time (duration in seconds of a stride), stride width (medio-lateral distance between heel position during heel strike of the left and right limbs). Hip-knee-ankle sagittal angle plots were used as visual representatives of intra-limb joint coordination. Mean and standard deviation of all kinematic data were obtained for each subject at every stage. Gait variability was assessed using standard deviation (SD) for the angular kinematic data, and coefficient of variation (CV) for all spatial temporal parameters. $CV = \frac{SD}{Mean} \times 100\%$

3.2.4 Data analysis

Student's t-test was used to identify between group differences in demographics, and fast-walking speed variables. Separated repeated analysis of variance using Bonferroni correction determined significant differences within-groups. If the assumption of sphericity was failed, a Greenhouse-Geiser correction was used. All statistical procedures were performed using SPSS 25 (IBM, NY, USA) and a significance level of 0.05 was used.

3.3 Results

The young and old adults achieved 70% of their maximum heart rate (intensity chosen for this study) at significantly different velocities (p = 0.010). Descriptive characteristics can be seen in Table 1. Old adults were not different from young adults in body mass, but were shorter in height than the younger counterpart.

Table 1: Characteristics of study's participants.

Characteristics	Young adults (n = 27)	Old adults (n = 23)
Age (years) *	26.6 ± 6.0	71.1 ± 5.6
Height (m) *	1.7 ± 0.1	1.6 ± 0.1
Body mass (kg)	68.0 ± 15.5	69.9 ± 10.2
Fast-walking speed (m/s) *	1.9 ± 0.2	1.2 ± 0.3

Note: Values are displayed as mean ± standard deviation.

* Significant difference (p < 0.05) between age-groups using independent sample *t*-tests.

Since gait strategies are directly influenced by the walking speed, both groups had differences in the cadence, stride length, stride time and step width mean values. During the activity both age-groups showed slight, but significant increase in stride length and decrease in cadence. Mean values of step width were kept constant throughout the task in both young and old adults, but the stride time increased only for the elderly group. Levels of variability (assessed by the coefficient of variation) were higher in the elderly group for the stride length and the stride time (see Figure 1). In both age-groups, statistically significant main effects of time were found only for the step width coefficient of variation (p = 0.03).

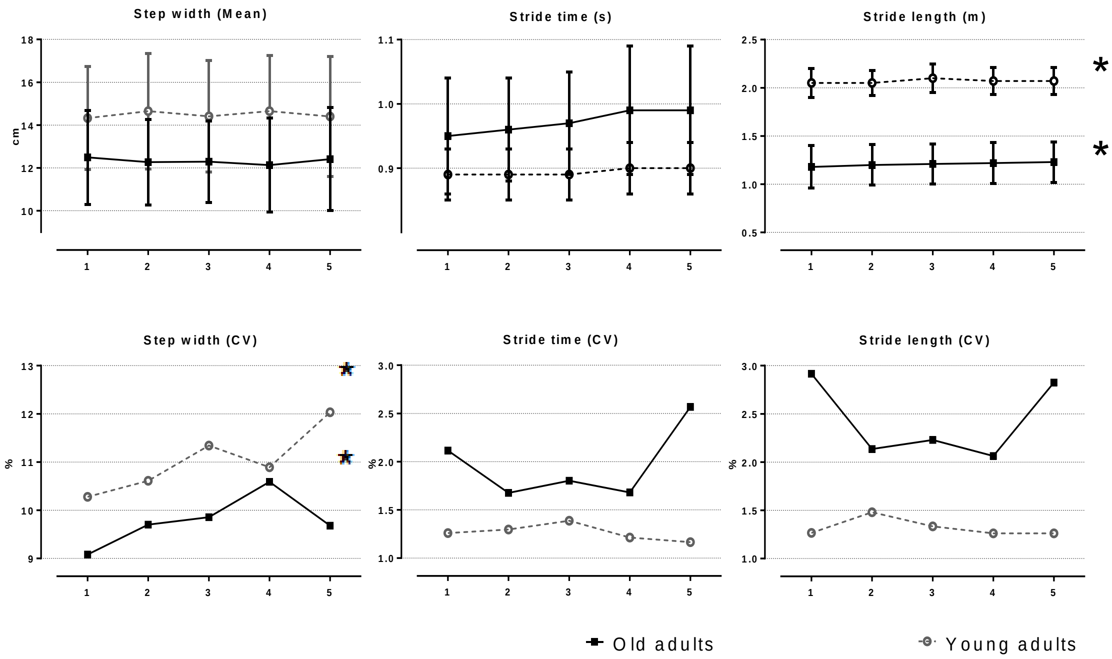

Figure 1: Spatial temporal parameters during fast-walking activity at the top. The mean and standard deviation for each subject were subsequently averaged across subjects within the age-group. At the bottom, differences in the coefficient of variation revealed the differences in spatial temporal variability along the task. * indicates statistically significant time effect.

Between-groups comparisons of joint angular kinematics showed differences for the mean values for the ankle, knee (at heel strike) and hip (at toe off). Significant effects of time were found for the ankle and hip at toe off for the elderly group, while for the young group, only the ankle at heel strike changed over time. Assessment of variability from the lower limb angular position (assessed by the standard deviation) showed old adults with higher variability at heel strike for the ankle and knee than the younger counterpart. No time effects were found for the lower-limb joint angles variability. Time and group differences of the sagittal angles at the heel strike and toe off is presented in Table 2, and can be seen in Figure 2.

Overall, the effect of the activity was more evident in the mean values of the lower limb angles, which can be evidenced by the difference in the intra-limb joint coordination patterns between the first and last (5th) stage of the protocol, as illustrated in Figure 3.

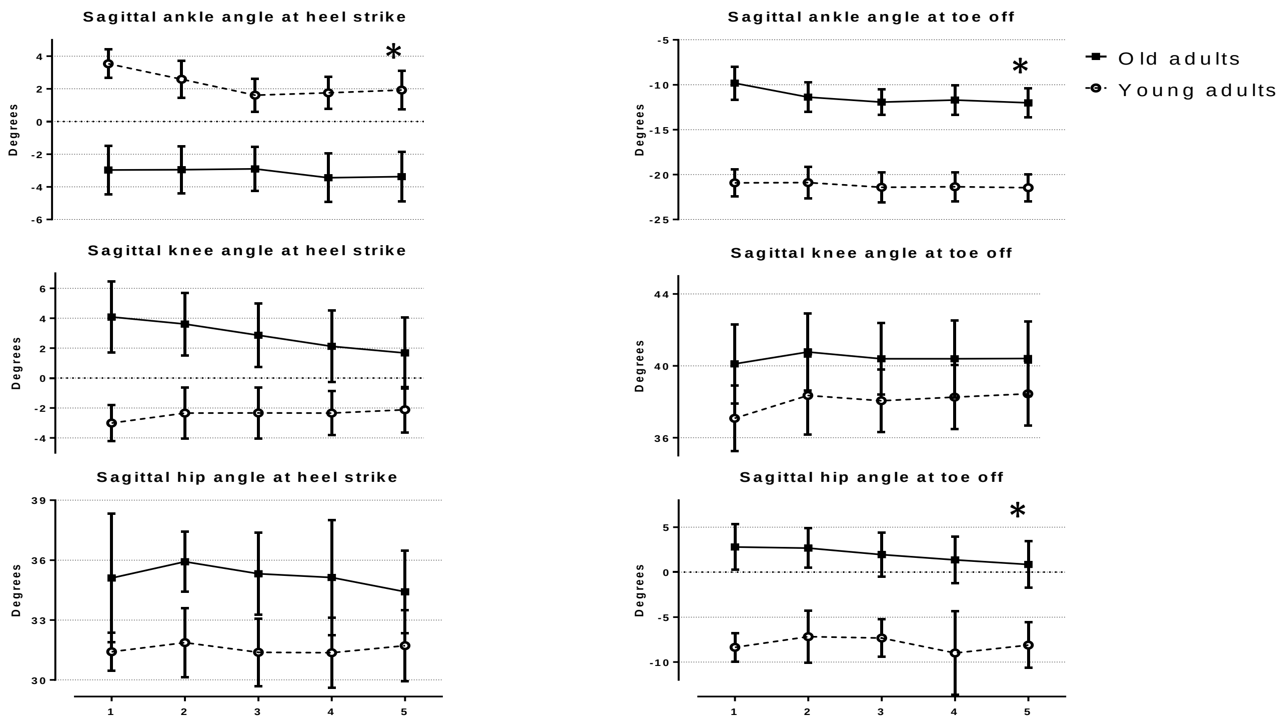

Figure 2: Angular displacements for the hip, knee and ankle at heel strike and toe off in young and old adults during the five time points along the fast-walking activity. * indicates statistically significant time effect. The mean and standard deviation for each subject were subsequently averaged across subjects within the age-group.

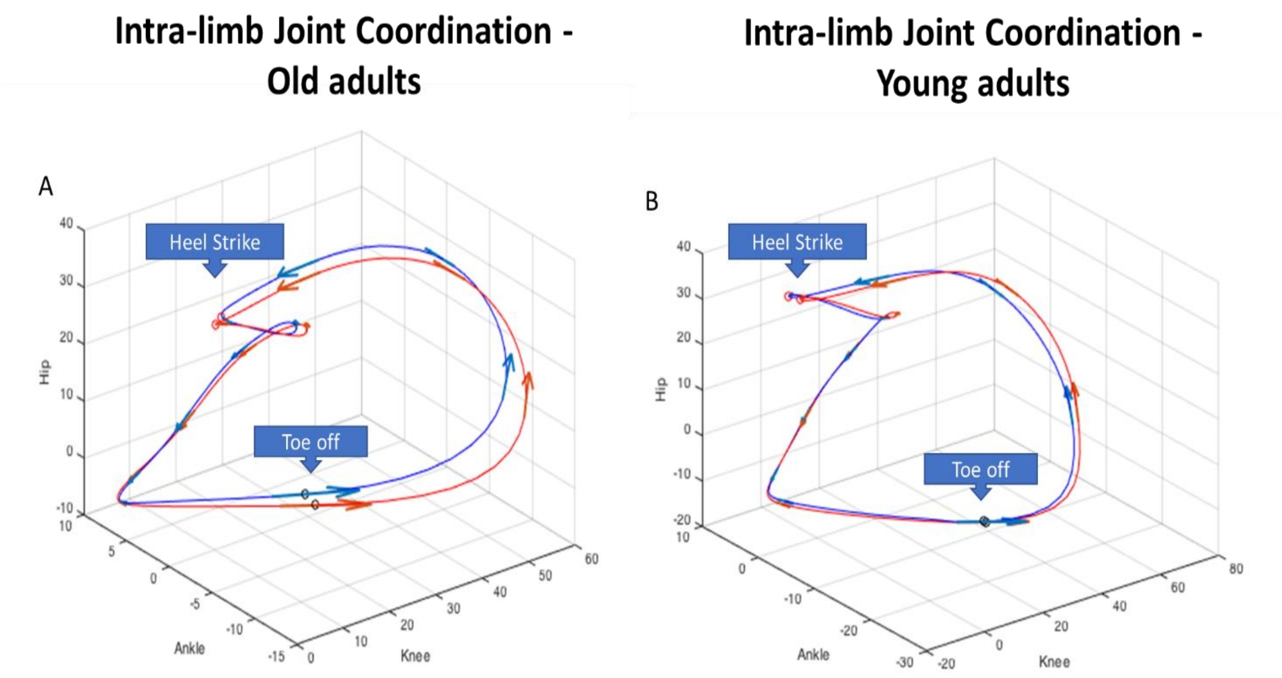

Figure 3: Illustration of Hip angle - Ankle angle - Knee angle intra-limb coordination patterns for the Old adults (A) and the Young adults (B). Blue line represents the mean values of the first stage, red line represents the mean values of the 5th / last stage. These two stages are displayed to highlight the differences induced by the activity. Arrows indicate the direction of the gait.

Table 2: Means and standard deviations for gait variables across all the time-points during fast-walking activity.

VARIABLE	GROUP	TIME POINTS					P ANOVA		
		T1	T2	T3	T4	T5	Group	Time	Interaction
Spatial temporal parameters									
Cadence	OA	**61.4 ± 0.8**	**62.2 ± 0.9**	**61.3 ± 0.8**	**60.2 ± 0.9**	**58.2 ± 0.9**	**.001**	**.004**	.518
	YA	**65.4 ± 0.8**	**63.8 ± 0.9**	**63.2 ± 0.8**	**62.2 ± 0.8**	**61.2 ± 0.7**			
Stride time (s)	OA	**0.95 ± .09**	**0.96 ± .08**	**0.97 ± .08**	**0.99 ± .10**	**0.99 ± .10**	**.022**	**.001**	**.020**
	YA	0.89 ± .04	0.89 ± .04	0.89 ± .04	0.90 ± .04	0.90 ± .04			
Step width (cm)	OA	12.5 ± 2.2	12.2 ± 2.0	12.2 ± 1.9	12.1 ± 2.2	12.2 ± 2.4	**.002**	.613	.226
	YA	14.3 ± 2.4	14.6 ± 2.7	14.4 ± 2.6	14.6 ± 2.6	14.1 ± 2.8			
Stride length (m)	OA	**1.18 ± .22**	**1.2 ± .21**	**1.21 ± .21**	**1.22 ± .21**	**1.23 ± .21**	**.001**	**.001**	.172
	YA	**2.05 ± .15**	**2.05 ± .13**	**2.1 ± .15**	**2.07 ± 14**	**2.07 ± .14**			
Stride length (CV)	OA	2.9 ± 2.6	2.1 ± 1.3	2.2 ± 1.3	2.1 ± 1.2	2.9 ± 3.8	**.012**	.341	.174
	YA	1.3 ± 0.4	1.5 ± 1.0	1.3 ± 0.7	1.3 ± 0.3	1.3 ± 0.3			
Stride time (CV)	OA	2.1 ± 1.8	1.7 ± 0.8	1.8 ± 1.2	1.7 ± 0.9	2.5 ± 1.4	**.007**	.377	.178
	YA	1.2 ± 0.4	1.3 ± 0.5	1.4 ± 1.0	1.2 ± 0.4	1.2 ± 0.3			
Step width (CV)	OA	9.6 ± 3.2	10.1 ± 3.6	10.4 ± 2.9	11.2 ± 3.8	11.6 ± 7.9	.319	**.031**	.402
	YA	11.5 ± 8.0	11.9 ± 8.7	12.6 ± 8.9	12.2 ± 8.5	13.2 ± 10.2			
Angular kinematics									
		Differences in the Mean Values along the activity							
Ankle angular	OA	-2.9 ± 5.5	-2.9 ± 5.0	-2.9 ± 4.6	-3.4 ± 4.8	-3.4 ± 5.0	**.001**	**.032**	**.034**
position at HS	YA	**3.5 ± 4.0**	**2.6 ± 5.0**	**1.3 ± 4.6**	**1.7 ± 4.0**	**1.9 ± 4.2**			
Ankle angular	OA	**-9.8 ± 6.0**	**-11.4 ± 6.4**	**-11.9 ± 6.4**	**-11.8 ± 7.0**	**-12 ± 6.5**	**.001**	**.021**	**.041**
position at TO	YA	-20.9 ± 10	-20.1 ± 11	-21.4 ± 10	-21.3 ± 10	-21.4 ± 10			
Knee angular	OA	4.0 ± 6.0	3.6 ± 6.6	2.9 ± 6.9	2.2 ± 6.5	1.7 ± 7.0	**.001**	.126	.101
position at HS	YA	-3.0 ± 3.4	-2.3 ± 3.4	-2.3 ± 3.2	-2.3 ± 3.4	-2.1 ± 3.2			
Knee angular	OA	40.1 ± 6.7	40.8 ± 6.4	40.4 ± 6.5	40.4 ± 7.1	40.4 ± 7.2	.127	.548	.374
position at TO	YA	37.1 ± 4.0	38.4 ± 4.1	38.0 ± 4.5	38.3 ± 4.4	38.4 ± 4.1			
Hip angle position	OA	35. 1 ± 7.6	35.9 ± 9.3	35.3 ± 9.4	35.1 ± 8.9	32.9 ± 10.8	.103	.279	.246
at HS	YA	31.4 ± 5.8	31.9 ± 6.2	31.4 ± 6.6	31.4 ± 6.0	31.7 ± 6.1			
Hip angle position	OA	**2.8 ± 9.9**	**2.7 ± 9.5**	**1.9 ± 9.8**	**1.4 ± 9.1**	**0.8 ± 9.4**	**.00**	**.017**	.032
at TO	YA	-8.4 ± 6.8	-7.1 ± 7.3	-7.4 ± 6.9	-9.0 ± 7.1	-8.1 ± 6.4			
		Differences in the Standard Deviation along the activity							
Ankle angular	OA	1.5 ± 0.8	1.4 ± 0.7	1.3 ± 0.9	1.5 ± 1.2	1.5 ± 1.2	**.013**	.657	.706
position at HS	YA	0.9 ± 03	1.1 ± 0.9	1.0 ± 0.4	1.0 ± 0.4	1.2 ± 0.7			
Ankle angular	OA	1.8 ± 0.9	1.6 ± 0.8	1.4 ± 0.7	1.6 =± 0.7	1.6 ± 0.8	.946	.562	.089
position at TO	YA	1.5 ± 0.5	1.7 ± 0.8	1.7 ± 0.6	1.6 ± 0.9	1.5 ± 0.6			
Knee angular	OA	2.4 ± 1.3	2.1 ± 1.1	2.1 ± 1.3	2.4 ± 2.3	2.4 ± 2.0	**.012**	.939	.326
position at HS	YA	1.2 ± 0.7	1.7 ± 1.2	1.7 ± 1.2	1.5 ± 1.0	1.5 ± 1.0			
Knee angular	OA	2.1 ± 0.7	2.1 ± 1.1	2 ± 1.1	2.1 ± 1.0	2.1 = 1.1	.159	.533	.684
position at TO	YA	1.8 ± 0.7	2.2 ± 2.0	1.7 ± 0.6	1.8 ± 0.9	1.7 ± 0.6			
Hip angle position	OA	3.2 ± 3.8	1.5 ± 0.9	2.0 ± 1.5	2.9 ± 3.4	2.0 ± 2.8	.078	.303	**.005**
at HS	YA	0.9 ± 0.3	1.7 ± 1.9	1.7 ± 1.1	1.8 ± 1.1	1.8 ± 1.4			
Hip angle position	OA	2.5 ± 1.6	2.2 ± 1.4	2.4 ± 2.0	2.6 ± 2.1	2.5 ± 2.4	.667	.344	.366
at TO	YA	1.6 ± 0.8	2.9 ± 4.4	2.1 ± 1.5	2.0 ± 2.1	2.5 ± 2.1			

Values are presented as mean ± standard deviation, **bold** - indicates statistically significant main effects of time (p < 0.05). OA= Old adults; YA= young adults; CV = Coefficient of variation; HS = Heel strike; TO = Toe off.

3.4 Discussion

The purpose here was to explore the acute effect of a sustained fast-walking along the activity on the gait parameters in healthy young and old adults. Therefore, exploratory information was searched regarding potential risks induced by a more demanding activity than walking at a freely chosen speed as in daily life activities. To recap, both age groups reached the target walking intensity at different speeds, then, the relative speed was kept the same for each of the participants, and constant throughout the test. Therefore, reasons other than walking speed were responsible for the alterations observed along the test.

Spatial temporal parameters of the gait can represent the individual gait strategy. In addition, previous studies showed that spatial temporal variables and their variability were associated with risk of falls (Hamacher et al., 2011; Hausdorff et al., 2001). In the present study both age-groups during the task had fewer but longer strides, such strategy may be an attempt to minimize energy cost (Kuo, 2001). On the other hand, such strategy may increase the risk of injury at the knee (Ardestani et al., 2016). Stride length and stride time variability were greater in old adults when compared to young participants herein. Similar findings were reported by Kang and Dingwell (2008b). Literature regarding age-related differences in gait variability (defined here as fluctuations in spatial temporal gait characteristics) and its relationship with walking speed are contradictory. Because old adults typically walk slower, increased gait variability in healthy old adults was considered to be related to their slow walking speed. However, previous studies have suggested that age-related changes in variability rather than be a manifestation of walking speed is more likely to reflect an underlying impairment of the motor system. (Hausdorff, 2007; Kang & Dingwell, 2008b; Ko et al., 2010).

Regarding changes in spatial temporal variability, the most relevant finding here was the effect of fast-walking activity over time on values of step width variability in both age-groups. Increased step-width variability has been associated with risk of falls (Maki, 1997; Toebes et al., 2012), loss of balance (Brach et al., 2005) and lowered postural control (Bauby & Kuo, 2000). Some authors suggest that increasing the variability of the step-width may be necessary to adjust the base of support and maintain the balance (Brach et al., 2005). Thus, during fast-walking activity, the increased step-variability could be a signal of incipient lateral instability (Brach et al., 2005; Gabell & Nayak, 1984).

The intra-limb coordination indicates the relative position between the joints. The coordination pattern reflects a characteristic of the motor control of organizing adjacent structures in terms of timing and position to execute a movement (Shumway-Cook & Woollacott, 2003). A given movement pattern reflects the control process required for spatial and temporal organization of different parts of the body (Magill & Anderson, 2007). Thus, to better understand the motor control process of a movement, the analysis of a single joint may not be sufficient. On the other hand, the intra-limb coordination pattern can display the coordinative synergism between the adjacent joints during a stride (Shafizadeh et al., 2013). Constraining the degrees of freedom in one joint would change this coordinative structure. Changes in the state of coordination may emerge due to alteration on any component of the motor system, as those caused by fatigue (Ferber & Pohl, 2011), or yet, alterations on muscle coactivation (Daly et al., 2010). As we observed from the intra-limb coordination trajectories, both groups displayed different patterns along the gait-cycle. In interpreting this finding, one must consider that this difference may be associated with differences in the walking speed between groups and not exclusively to the age-related differences in walking pattern. The differences from the 1st to the 5th (last) intra-limb coordination trajectories addresses our main objective. The shift between the two trajectories reflect the effect of the activity on the relationship between the joints throughout the gait cycle. Moreover, the differences between the intra-limb coordination trajectories (patterns) suggest greater differences in motor strategy during the swing phase when compared with the stance phase, especially for the old group.

Fast-walking resulted in significant changes on the angular kinematics of both age-groups. Towards the end of the practice, old adults were initiating the swing phase with additional ankle plantarflexion, and less extended hip. A reduction in hip extension may imply a functional tightness that would be preventing the hip to fully extend at the toe off (Lee et al., 1997). The increased ankle plantarflexion could be a compensatory mechanism to the reduction in hip extension since they managed to keep and even increase the stride length (JudgeRoy et al., 1996). These findings suggest that the relative contribution of the ankle plantar-flexion for propelling the body forward has increased during the activity, which may be a manifestation of a redistribution of joint torques and their relative contributions to the total performance (Lewis & Ferris). Meanwhile, younger adults manifested early signs of muscle fatigue, as we can see through their progressive reduction of the ankle dorsiflexion at landing (Parijat & Lockhart, 2008).

In respect of kinematic variability, old adults displayed higher variability for the ankle and knee angular position at heel contact, but, levels of kinematic variability did not change throughout the protocol. However, in the present study, analysis regarding variability of angular kinematics was restricted to two events of the gait cycle. Further study on progressive effects of fast-walking activity should assess kinematical variability throughout the gait cycle to provide a complete understanding of the effects of fast-walking.

To the best of our knowledge, this study was the first to describe alterations on gait during sustained walking at a faster pace. Our findings revealed that both age groups changed the gait strategy by reducing cadence and increasing stride length, while progressively increased the variability of step width. Our analysis on angular kinematics revealed differences throughout the activity. Alterations found for the young group were compatible with that of fatigue installation, whilst the old group seemed to change the power generation strategy for propelling the body forward.

Walking at faster pace had no effect on angular kinematics variability assessed at the heel strike and toe off. Further analysis should verify the effect of fast-walking throughout the entire gait cycle. Differences in intra-limb coordination trajectories revealed that the effect of fast-walking on the pattern of coordination between the lower limb joints were more prominent during the swing phase. This finding may indicate an increment in the risk of falls by tripping or slipping since the swing foot trajectory is sensitive to the coordinated movements of the lower-limb joints.

Most of studies investigating effects of speed on gait have not assessed the progressive acute effects induced by sustained fast-walking. Our findings revealed that old adults progressively changed their kinematics at hip and ankle at the beginning of the swing phase. Meanwhile, young adults showed incipient signs of fatigue at the ankle joint. In both age groups, the effect of the activity in the coordination of lower limb angles were more evident during the swing phase. Future work should investigate kinematic alteration throughout the gait cycle, particularly at the swing phase. The findings reported herein are important for addressing potential risks associated to walking at a faster pace.

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Conflict of interest

None of the authors have any conflicts of interest associated with this study.

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Chapter 4

Effects of Fast-Walking on Muscle Activation in Young Adults and Elderly Persons

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Abstract

Coactivation of agonist and antagonist muscles participates in the regulation of joint stiffness and postural instability. Alterations on muscle activity have been revealed as an important falling risk factor. It is unclear the effects, and age-related differences, of a prolonged demanding task on the muscular coactivation levels. We compared muscle activation amplitude and coactivation of the vastus medialis, biceps femoris, tibialis anterior, and gastrocnemius medialis from surface EMG in 16 young adults (age 21-33) and 8 elderly adults (age 66-72) while fast-walking at 70% of their maximum heart rate. Overall, the elderly demonstrated higher coactivation indexes than the young individuals. Ankle coactivation decreased in the first half of the swing phase, while coactivation at the knee increased in the latter half of the swing phase in our elders. Alterations of muscle activation and coactivation on the knee and ankle were more prominent close to landing and during the swing phase. Our results suggest that these alterations may suggest potential concerns with respect to the risk of falls.

KEYWORDS: COACTIVATION; ELECTROMYOGRAPHY; FALL IN ELDERLY; FATIGUE; ACTIVITIES OF DAILY LIVING

4.1 Introduction

The ageing of the population is a worldwide phenomenon. In 2050, Europe will have an elderly population of 31.9%, i.e. one in each three people will be over the age of 65 (Eurostat, 2015). This demographic behavior has led to new challenges in public health given the financial impact caused by the number of hospitalizations of this population, most of them due to consequences of falls. Risk factors of falls in elders are usually related to the unsuccessful interactions between environmental demands and intrinsic factors (Tinetti & Speechley, 1989). Age-related decrements in neuromuscular function, control of balance and strength, are all key factors that when isolated or combined make the elderly more susceptible to falls (Gandevia, 1998; Tinetti & Speechley, 1989). Since most of falls occur during locomotion, decrements in dynamic postural control and gait performance has been pointed as risk factors for falling (Lee & Kerrigan, 1999).

Stable walking kinematics requires generating appropriate motor patterns, in which muscle coactivation plays an important role (Lee & Kerrigan, 1999). In a variety of activities, including walking, elderly use more coactivation than young adults (Hortobagyi & DeVita, 2000). When walking at a faster pace, coactivation of antagonist leg muscles has shown to be greater in old adults and to have a high association to metabolic rate (Peterson & Martin, 2010). Thus, an activity that is more energy consuming, such as fast walking, which implies higher coactivation levels, may compromise gait performance over time in old adults. According to Pereira and Gonçalves (2011a), prolonged walking on a treadmill induced changes in the lower limb muscle activities, however how the muscle activation patterns changes over time during fast walking were not described. Although a prior study regarding effects of fast-walking on the pace of elders suggested that it was safe to perform fast-speed walking exercises (Fan et al., 2016). Nagano et al. (2014) demonstrated that after a short period of a fast walk older individuals were more susceptible to falls by tripping. Nevertheless, it remains unclear whether fast-walking can be performed without imposing risk to the elderly.

Coactivation of agonist and antagonist muscles participates in the regulation of joint stiffness and postural instability (Hortobagyi & DeVita, 2000;

Schmitz et al., 2009). Thus, alterations on muscle activity may reflect walking stability. To understand fast-walking effects on gait performance, it is of interest to analyze the neuromuscular alteration over time.

The purpose of this study was to investigate the influence of fast-walking on activation and coactivation of lower limb muscles of young and old adults. We hypothesized that old adults would have greater muscle activation and coactivation than the younger individuals during a gait, and that the levels of muscle activation and coactivation would increase throughout the task.

4.2 Methods

4.2.1 Participants

Data of 15 adults (27 ± 5 years; 1.68 ± 0.12 m; 68.3 ± 17.3 Kg) and 8 elderly persons (69.6 ± 3.8 years; 1.56 ± 0.1 m; 66.9 ± 8.2 kg) were considered for this study. Adults were recruited from among the students of the University and elderly participants were recruited from the local community to engage in this study. The subjects are described in Table 1. The exclusion criteria were any orthopedic, neurological, visual, vestibular or cardiovascular conditions that would not allow the subject to perform all of the proposed activities. All participants gave signed consent to participate in the study. This research was approved by the Local Ethics Committee.

Table 1: Subject characteristics.

	Old adults	Young adults	*p*-value
Age (years)	69.9 ± 3.8	27.5 ± 5.1	< 0.001
Walking speed (m/s) [a]	1.3 ± 1.0	1.9 ± 2.0	< .01
Body mass (kg)	68.0 ± 8.2	68.3 ± 17.2	0.39
Height (m)	1.6 ±1.0	1.7 ± 1.0	0.19

Notes: Values are expressed as mean ± SD and median; [*]independent *t*-test; [a]Average fast-walking speed of the task.

4.2.2 Task and Procedures

All participants completed a familiarization session on an treadmill (AMTI Inc, MA, USA) walking at their preferred walking speed for five minutes. Elderly men and women wore a safety harness attached to the ceiling. After a familiarization period of 5 minutes, the treadmill speed was increased by 0.5 km/h every 30 seconds until the participant reached the speed where they would be at 70% of their age-predicted maximal heart rate (220 minus age in years). The participants were instructed to walk at this intensity for twenty minutes or until the voluntary exhaustion. Heart rate (HR) was measured continuously during the test with a Polar HR monitor (2010 Polar Electro Oy, FI-90440 Kempele, Finland) to ensure they were walking at the required intensity. Additionally, the rating of perceived exertion scale was obtained throughout the protocol using a modified Borg 10-point scale (Borg, 1982).

Data were recorded for 30 seconds every minute until the end of the protocol. Since not all the participants were able to walk during twenty minutes, five time points equally distributed were used to analyze the data over time. The first and the last thirty seconds of the recorded data, the second time point was the 30s recorded at the middle point between the first and the last, and the second and fourth time point from the middle time between the first and the third time point, and the third and last time point, respectively.

Electromyography of the right limb's vastus medialis (VM), biceps femoris (BF), tibialis anterior (TA) and gastrocnemius medialis (GM) muscles were recorded using a Trigno wireless acquisition system (Delsys Inc., Boston, MA, USA) at a sampling rate of 200Hz, according to SENIAM guidelines. Surface electrodes with an inter-electrode distance of 10 mm, 4-bar formation, and bandwidth of 20–450 Hz, were applied to the midline of the palpated belly muscle, parallel to the muscle fiber orientation. Recording sites were shaved and cleaned with alcohol and then the electrodes were affixed to the skin with hypoallergenic tape. The electrodes were further secured to reduce motion artifact using flexible, non-adhesive wrap encircling the thighs (Retelast™ Tubular Stretch Bandage Large, 3M Healthcare). Both EMG and kinematics was recorded using an infrared motion capture system (Qualisys AB, Gothenburg, Sweden). Gait events for heel strike and toe-off were computed using only kinematic data based on the heel and toe markers trajectories (Zeni et al., 2008). The EMG data were processed to extract the average linear envelope during gait cycle. EMG signals were bandpass filtered (Butterworth digital, fourth order, 20 Hz - 500 Hz), rectified and the RMS of the signal was obtained using 100ms moving continuous windows

(Visual 3D v.5, C-motion Inc., Rockville, MD, USA). RMS EMG was quantified and averaged over 30 consecutive cycles of the following phases: absorptive phase (from heel strike to peak knee flexion), and propulsive phase (from peak knee flexion to toe off). All data were normalized and expressed as a percentage of the maximal EMG detected during the first time point of the task, which corresponds to the first 30 seconds of the trial. The EMG signal were time normalized to the gait cycle starting from heel strike of the right foot to the subsequent ipsilateral heel strike. The average of 30 gait cycles provided one representative curve for each participant muscle activation data. Muscle amplitude data was ensembled averaged to obtain one curve per muscle for all participants at the five time points.

The magnitude of MG-TA and BF-VM coactivations were quantified with coactivation index and calculated across the loading response portion of the gait, from 0 to 25% of the stance phase; the midstance and propulsive phase of support, from 25 to 100% of the stance phase; and the first and second half of the swing phase. The coactivation index (CoI) was calculated using the following equation:

$$\mathbf{CoI} = \frac{\int \boldsymbol{EMG_{ANT}}}{\int \boldsymbol{EMG_{ANT}} + \boldsymbol{EMG_{AG}}} \times \mathbf{100}\%$$

Where $\boldsymbol{EMG_{ANT}}$ is the magnitude of EMG from the lower muscle activity, and $\boldsymbol{EMG_{AG}}$ is the magnitude of EMG from the higher muscle activity (Kellis et al., 2003). Respective values of CoI were obtained from the averaged values for each participant and submitted for statistical analysis (see Figure 1).

4.2.3 Data analysis

Data is reported as means ± standard deviations. Separate mixed model repeated measures ANOVAs (2 x 5) were used to detect differences between age-groups and within groups for time and interaction effects for all the dependent variables. All analyses were performed in SPSS (version 24, IBM SPSS, Chicago, IL.), with significance accepted at $p < 0.05$.

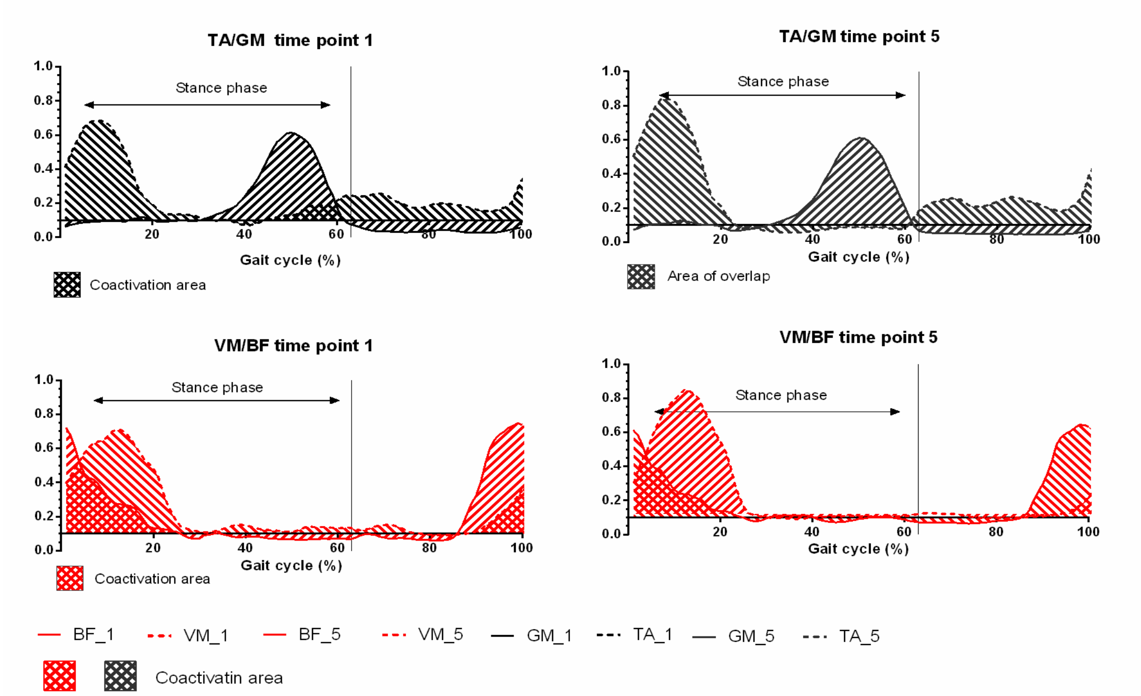

Figure 1: Illustration of coactivation area. Lines represent the mean linear envelope of muscle 1 activation. Dashed lines represent the mean linear envelope of muscle 2 activation. Crosshatched areas represent regions of coactivation.

4.3 Results

Normalized RMS EMG amplitude during the first part of the stance phase (0-25%) were not different in young adults in comparison to the older people. The task protocol significantly increased RMS EMG of TA in the old adults over the absorptive phase, $F(1.7, 31.4) = 4.66$, $p = 0.021$, with significant differences between the elderly and the young subjects, $F(1, 18) = 5.81$, $p = .027$. For the rest of the support phase (25 to 100% of the stance phase), RMS activity for the TA decreased to a similar extent for both age groups, $F(4,72) = 4.230$, $p = 0.004$, (Fig.2). Overall, the elderly demonstrated higher coactivation indexes than the younger subjects (Fig.3). There was no significant time effect CoI during the absorptive phase (Fig. 3a, 3e). Meanwhile, coactivation between VM and BF in the adults during the midstance and propulsive phase of support decreased significatively relative to the elderly, $F(4,64) = 3.164$, $p = 0.02$, with significant difference between groups, $F(1,16) = 7.192$, $p = .016$ (Fig. 3f). Interaction effects were not significant. During the first half of the swing phase, young adults had no coactivation between the pair of muscles. The analysis was computed only for the old adults. During this phase, CoI between TA and GM decreased significantly (p = 0.02), whilst coactivation between VM and BF increased (p = 0.01) (Fig 3c, 3g). During the second half of the swing phase, CoI between VM and BF significantly decreased in both groups, $F(2.1,37) = 5.274$, p =.009, (Fig. 3h).

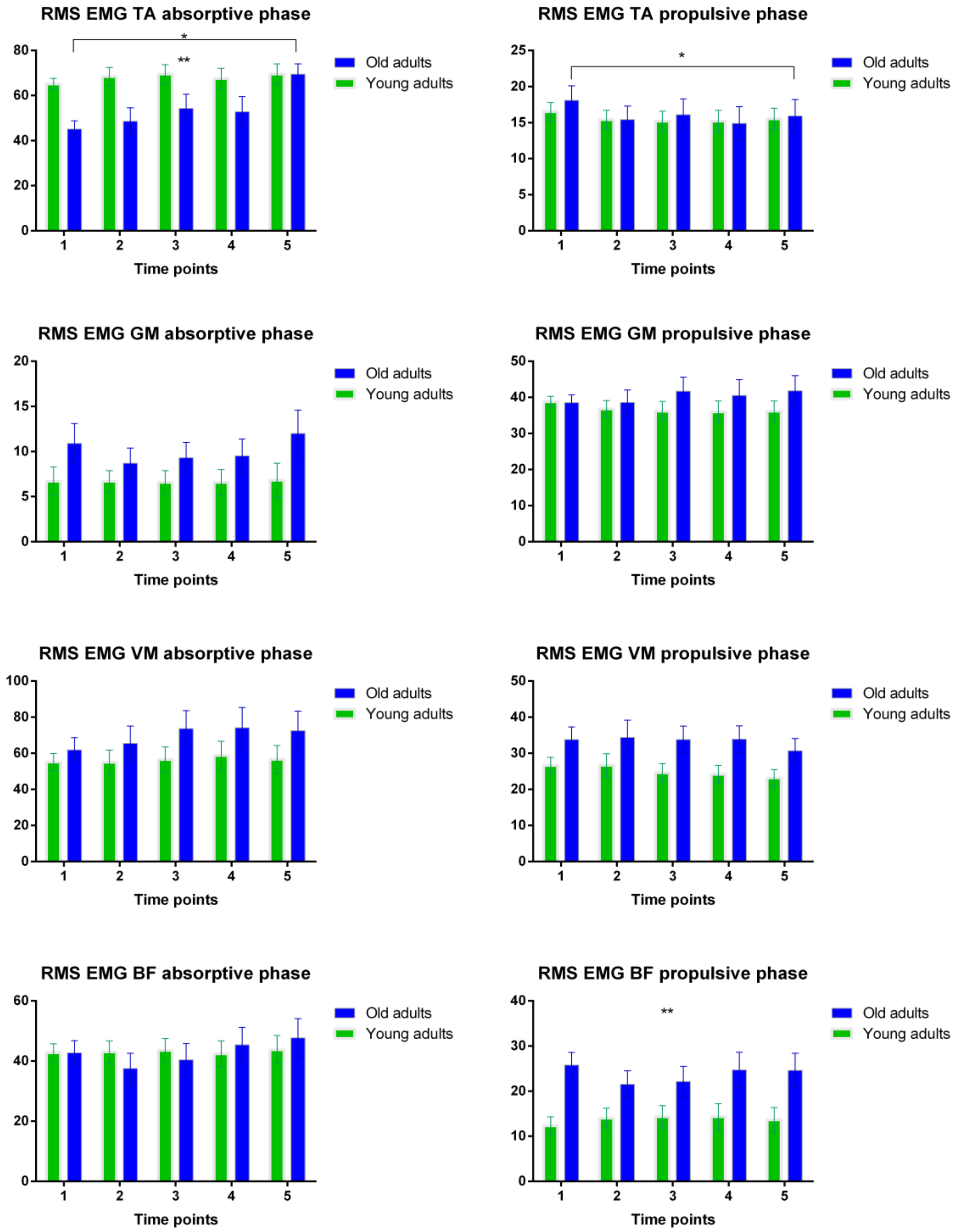

Figure 2: The changes of muscle activities in Vastus Medialis (VM), Biceps Femoris (BF), Tibialis Anterior (TA), and Gastrocnemius Medialis (MG) for young and old adults over time. * represents significant changes over time; ** represents significant differences between groups.

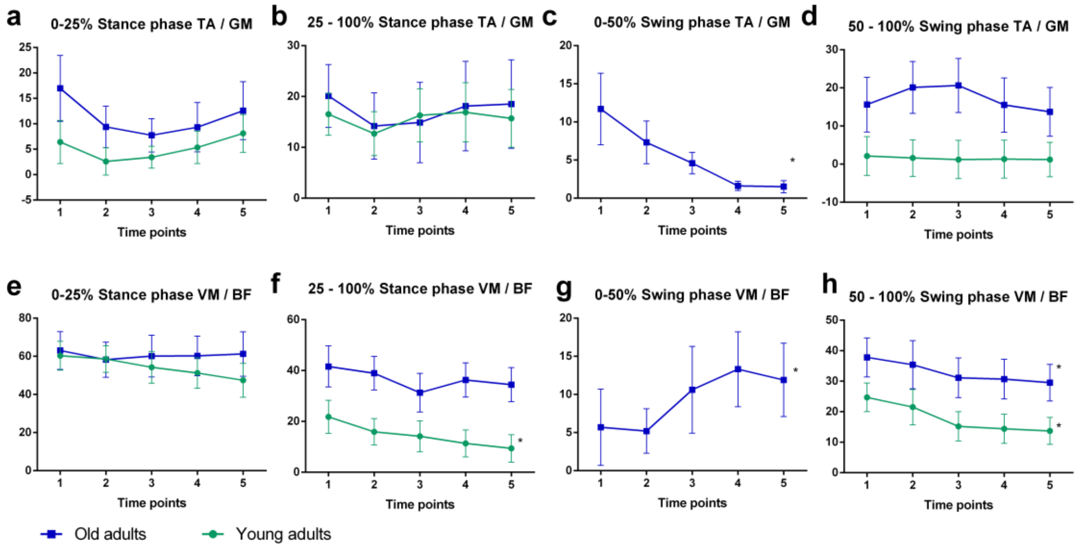

Figure 3: Changes during the activity of the coactivation Indices between older (blue line) and young adults (green line). * represents significant differences over time.

4.4 Discussion

We hypothesized that old adults would demonstrate increased muscle activation and coactivation when compared with younger adults. Overall, in the present study, we found that elderly individuals exhibited higher levels of muscle activation and coactivation about the thigh and shank than young adults, which is consistent with several previous studies (Hortobágyi et al., 2009; Mian et al., 2006; Peterson & Martin, 2010). Higher muscle coactivation in old adults is expected because of the age-related reduced stretch-reflex excitability imposed by central and peripheral processes (Obata et al., 2010). We also hypothesized that old adults would demonstrate increased muscle activity and increments in the magnitude of agonist and antagonist muscle coactivation throughout the task. In support of this hypothesis, we found significant increases in tibialis activity during the absorptive phase. However, in general, muscle coactivation decreased during fast walking in both age groups. Excepted by the significant increment in the coactivation between vastus medialis and biceps femoris during the first half of the swing phase in old adults.

Fatigue effect on muscle coactivation is contradictory and it seems to rely on the type of task. An isotonic fatigue protocol did not alter muscle coactivation of the knee flexors or extensors (Longpré et al., 2013). Although, increased ankle coactivation were detected after an isokinetic fatigue protocol (Granacher et al., 2010). Meanwhile, fatigue induced by cyclic activities, such as walking and cycling, reduced coactivation levels (Hautier et al., 2000; Pereira & Gonçalves, 2011a). Indeed, alterations in coactivation levels in the knees and ankles exhibited during fast-walking in the present study were similar to those previously found in Pereira and Gonçalves (2011a). Finally, the overall reduction in muscle coactivation levels observed in the present study may have been caused by the prolonged time walking at a faster pace, and ultimately to fatigue.

Successful walking kinematics requires generating appropriate motor patterns. Muscle coactivation plays an important role in the coordination of the lower limbs in elders throughout the gait cycle (Lee & Kerrigan, 1999). Therefore, we address the interpretation of results regarding the role of muscle activation and coactivation within the gait cycle. The reduction of coactivation levels at the knee during terminal swing in the present study has been suggested to be a safety motor strategy (Pereira & Gonçalves, 2011b). At this phase of the gait, the swing leg is moving forward in preparation for load acceptance, an unexpected disturbance at this point requires a fast response from the knee, higher levels of coactivation may affect the capacity of the knee extensors to produce force in a short period of time (Parijat & Lockhart, 2008). On the other hand, the reduction of coactivation at the knee, reported in the present study, may have been the result of fatigue caused by the exertion of the activity, which may cause a delay response in producing knee joint moment. Previous studies suggest that old adults may adopt a strategy of redistributing the total available torque by using proportionally less knee torque and more hip torque (Hortobágyi et al., 2003). In this case, further analysis should verify if fast-walking compromises hip performance, since a distal to proximal shift in joint kinetics requires greater efforts of the proximal muscle extensors.

Interestingly, most of the alterations in coactivation level were found in the swing phase. Ankle coactivation reduced significantly in the first half of the swing phase, but maintained its levels in the latter half. While knee coactivation significantly increased in the first half of the swing phase, it subsequently decreased in the second half of the swing phase. The principal swing phase task is the progression of the foot of the swing limb from the previous to the next support position (Winter, 1992). Risk of falls by tripping is strictly associated to the swing toe trajectory, which is especially sensitive to the coordinated movement of the lower limb joints. In previous work, Nagano et al. (2014) have demonstrated the potential risk of tripping following a period of fast-walking activity. The alteration on coactivation in the ankle and knee reported herein may affect the mechanical efficiency of the swing limb, therefore affecting toe trajectory. The reduction in ankle coactivation in the first half of the swing phase throughout the task could have been caused by fatigue or yet, due to an increment in the dorsiflexor activity to ensure safety values of the toe height.

Changes in muscle activation and coactivation emerged during the task, which could ultimately affect the coordinated motion of the lower limb segments. Additionally, fatiguing of the lower extremity musculature has been suggested to influence gait variables pertinent to slip initiation and recovery phase (Parijat & Lockhart, 2008). Alterations in knee and ankle muscle activation due to fatigue were more prominent in specific sub phases of the gait, such as landing and swing phase. Our results suggest that these alterations may accommodate potential concerns with respect to risk of falls. Some limitations of this study were the absence of kinetic analysis and the small sample size. Consequently, the findings presented in this paper provide a baseline for further research and may be used in studies regarding the effect of fast-walking on muscle activation. Further studies could extend the analysis of discrete parameters to intra-stride analysis, which could provide more information regarding the fatigue effects on muscle activity within the gait cycle.

In summary, fast-walking induced neuromuscular adaptations consistent with previously reported studies based on fatigue. Most of the alterations found in the present study occurred at sub phases of gait associated to critical events that may increase risk of falls by tripping or slipping. It is therefore of interest to perform subsequent analysis regarding the biomechanical alterations that may be caused by fast-walking activity.

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Conflict of interest

The authors of this paper are not in any conflict of interest with regard to the work presented.

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Chapter 5

Age-related changes in kinematic and muscle activation variability during fast-walking

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Paper submitted for publication

Abstract

Walking at a faster pace has been found to increase gait variability increasing risk of falls in old adults. To further explore this phenomenon, the aim of this study was to assess the variability of lower limb muscles' activity and kinematics in younger (n = 15, age = 27 ± 5 years) and old adults (n = 8, age = 70 ± 4 years) walking at faster pace. Variability was assessed based on the mean standard deviation during the gait cycle, where values higher than the mean indicated regions with higher variability. Analyses of variance were used to assess differences in variability between age groups and over time. The tibialis anterior (TA) activation variability was higher among the old adults ($p = 0.01$). Fast-walking reduced TA interstride variability ($p = 0.03$) in both groups, and it reduced the gastrocnemius medialis variability in the old adults ($p<0.001$). Thus, the muscle activation pattern was more sensitive to perturbations in old adults along the task. Despite the muscle activation variability differences found, there were no significant differences in the lower limb kinematics variability between the age groups or over time. The intra-stride variability assessment revealed different patterns of variability within the gait cycle between the groups. Changes in intra-stride variability seems to suggest a shift in the motor strategy during prolonged walking, and could reveal increased variability in portions of the gait cycle associated to increased risk of falls by tripping or slipping. However, this needs to be assessed in future studies.

KEYWORDS: OLD ADULTS, FALLS RISK, GAIT VARIABILITY, ELECTROMYOGRAPHY, KINEMATICS.

5.1 Introduction

Falls are very common in old adults, and one third of these falls result in serious injury (Hausdorff et al., 2001). Most consequences of falls are serious injury, hospitalization and physical dependency (Rubenstein, 2006). Early prevention is, therefore, essential so that falls and, ultimately, their consequences, can be prevented. A large portion of falls occur during walking, and tripping or slip-induced accidents are amongst the most common causes (Niino et al., 2000). Thus, considerable efforts have been directed towards the maintenance of stability on human walking. To facilitate walking stability, humans must generate appropriate motor patterns and effective corrective response to perturbations. A general muscular exercise, such as fast-walking, places greater demand on physiological systems in old adults (Peterson & Martin, 2010). In addition, previous studies have shown that fast-walking increases gait variability and compromises gait performance (Callisaya et al., 2011; Nagano et al., 2014).

Gait variability is resultant of interstride variations. Repetitive/cyclic motor tasks are executed within some degree of variability, reflecting the adaptability of the system to perturbations (Stergiou & Decker, 2011). However, high variability has been related to increased risk of falls because it can reflect less effective motor system corrections. Gait variability can be used to assess gait stability and it can be influenced by walking speed (Hamacher et al., 2011). The impact of fast-walking on gait stability can be assessed based on gait variability, which reflects the strategies of the motor system in response to walking at a faster pace. The objective of this study was to assess the adaptive strategies used by the motor systems of younger and old adults in response to walking at a faster pace by evaluating the variability of lower limb muscles' activity and kinematics.

5.2 Methods

5.2.1 Participants

The participants were 15 young (27 ± 5 years; 1.68 ± 0.12 m; 68.3 ± 17.3 Kg) and 8 old adults (70 ± 4 years; 1.56 ± 0.1 m; 66.9 ± 8.2 kg). Younger adults were recruited from students of the University and old adults were recruited from the local community. The exclusion criteria were any orthopedic, neurological, visual, vestibular or cardiovascular conditions that would not allow the subject to walk at a faster pace for a prolonged time. All participants signed an informed consent form before participation. This research was approved by the Local Ethics Committee.

5.2.2 Task and Procedures

All participants completed a familiarization session on a treadmill (AMTI, Inc. MA, USA) by walking at their preferred walking speed for five minutes. The older participants wore a safety harness attached to the ceiling to prevent falls. After familiarization, the treadmill speed was increased by 0.5 km/h every 30 seconds until the participants reached 70% of their maximal heart rate (220 - years of age). The participants walked at this intensity for twenty minutes or until exhaustion. Heart rate (HR) was measured continuously during the test with a Polar HR monitor (2010 Polar Electro Oy, FI-90440 Kempele, Finland). The perceived exertion was assessed during the test using the modified 10-point Borg scale (Borg, 1982).

Prior to data acquisition, participants were prepared for surface electromyography (sEMG) recording. The electrode sites were shaved and cleaned with alcohol; then the electrodes were affixed to the skin with hypoallergenic tape following the SENIAM guidelines (reference). Surface electrodes had an inter-electrode distance of 10 mm, 4-bar formation, and bandwidth of 20–450 Hz. They were applied to the midline of the muscle bellies, parallel to the muscle fibers orientation of the vastus medialis (VM), biceps femoris (BF), tibialis anterior (TA) and gastrocnemius medialis (GM). The electrodes were further secured to reduce motion artifact using flexible, non-adhesive wrap encircling the thighs (Retelast™ Tubular Stretch Bandage Large, 3M Healthcare). sEMG was recorded using a Trigno wireless acquisition system (Delsys Inc., Boston, MA, USA) at a sampling rate of 1000Hz.

The gait parameters were recorded using an 8 Oqus cameras motion capture system (Qualisys, Inc., Gothenberg, Sweden) operating at 100 Hz. Reflective markers were placed bilaterally on the posterior and anterior superior iliac spines, greater trochanter, medial and lateral femoral condyles, medial and lateral malleoli, posterior heels, first metatarsal heads. Kinematics and sEMG were recorded simultaneously for 30 seconds every minute until the end of the task. Five time points equally distributed as a percentage of the total walking time were used to analyze changes over time.

Kinematics and sEMG data were analyzed in Visual 3D (v.5, C-motion, USA). Trajectories were smoothed with a fourth-order zero-lag low-pass Butterworth filter with a 12 Hz cut-off frequency. Heel strike was determined as the maximum anterior position of the heel marker relative to the pelvis. A gait cycle was defined by consecutive heel strikes of the right foot. Joint angles were calculated using Cardan sequence of rotations (x–y–z), where sagittal joint flexion occurred about the x-axis. Positive values denoted flexion. Sagittal joint angles at the ankle, knee and hip, its respective angular velocities and sEMG were time normalized to 100% of the gait cycle. sEMG signals were bandpass filtered (Butterworth digital, fourth order, 20 Hz - 500 Hz), rectified and the RMS of the signal was obtained using 100 ms windows. All sEMG waveforms were amplitude normalized to the maximum amplitude obtained for each muscle during the first 30 seconds of the trial. For every time-point, the mean of 30 gait cycles provided one representative curve for each variable. Means and standard deviations of the joint angles, velocities and sEMG linear envelopes were calculated at each percentage of the gait cycle. To determine the variability of these measures, at each percent of the gait cycle, standard deviations among strides were calculated, and then averaged over the gait cycle (MeanSD), according to the formula below (Bruijn et al., 2009; Dingwell & Cavanagh, 2001; Dingwell & Marin, 2006; Kang & Dingwell, 2008):

$$\boldsymbol{MeanSD} = \langle \boldsymbol{SD(i)} \rangle\, \boldsymbol{i}, \boldsymbol{i} \in \{\mathbf{0}\text{–}\ \mathbf{100}\ \%\ \boldsymbol{gait\ cycle}\}$$

where SD(*i*) indicates the standard deviation of a measure at *i*th % gait cycle, and ⟨ - ⟩*i* denotes the average over all *i* (time-points). Thus, for each subject the MeanSD characterized the variability for the entire gait cycle of each parameter of interest.

Intra-stride variability was quantified by extending this analysis of MeanSD. First, we established the MeanSD for each dependent variable. Second, we identified every ith % point higher than the MeanSD. Then, we accounted for the total of the sample to allow the visualization of regions with higher variability within the gait cycle among the subjects. Kinematic and sEMG inter-stride variability (MeanSD's) were compared between age groups and over time using a general linear model analysis of variance (ANOVA) in SPSS 21 (IBM SPSS, Chicago, IL).

5.3 Results

There were significant differences between younger and old adults on the average fast-walking speeds that made them reach 70% of their maximum heart rate (1.9 ± 2 m/s vs. 1.3 ± 1 m/s, $p<0.01$). The effects of time and age on kinematics and sEMG variability are presented in Table 1. The TA sEMG variability was larger in the old adults ($p=0.01$), but the variability decreased over time for the TA in both groups ($p=0.03$), and also for the GM ($p<0.001$) in the old adults. Despite the muscle activation variability differences found, there were no significant differences in the lower limb kinematics variability between the age groups or over time.

Table 1: Effects of time and age on kinematics and EMG variability (*p*-values).

Interstride variability (MeanSD)	Time	Age	Time X Age
Vastus medialis	0.46	0.65	0.52
Biceps femoris	0.27	0.26	0.40
Gastrocnemius medialis	**0.001**	0.21	0.48
Tibialis anterior	**0.03**	**0.01**	**0.04**
Ankle angular displacement	0.40	0.31	0.39
Ankle angular velocity	0.12	0.32	0.69
Knee angular displacement	0.65	0.16	0.71
Knee angular velocity	0.92	0.26	0.40
Hip angular displacement	0.19	0.31	0.37
Hip angular velocity	0.81	0.65	0.94

Notes: A *p*-value in bold type denotes a significant difference.

The younger adults demonstrated an overall reduction in the variability within the gait cycle of the angular joints displacements. For the old group the intra-stride variability of the knee angular displacement reduced in the last time-point, particularly during the swing phase. Meanwhile, the intra-cycle gradient of variability (region with higher variability among the subjects) of the hip and ankle angular displacement seems to fade away when compared with the first time point, particularly in the stance phase (Figure 1). The intra-stride variability

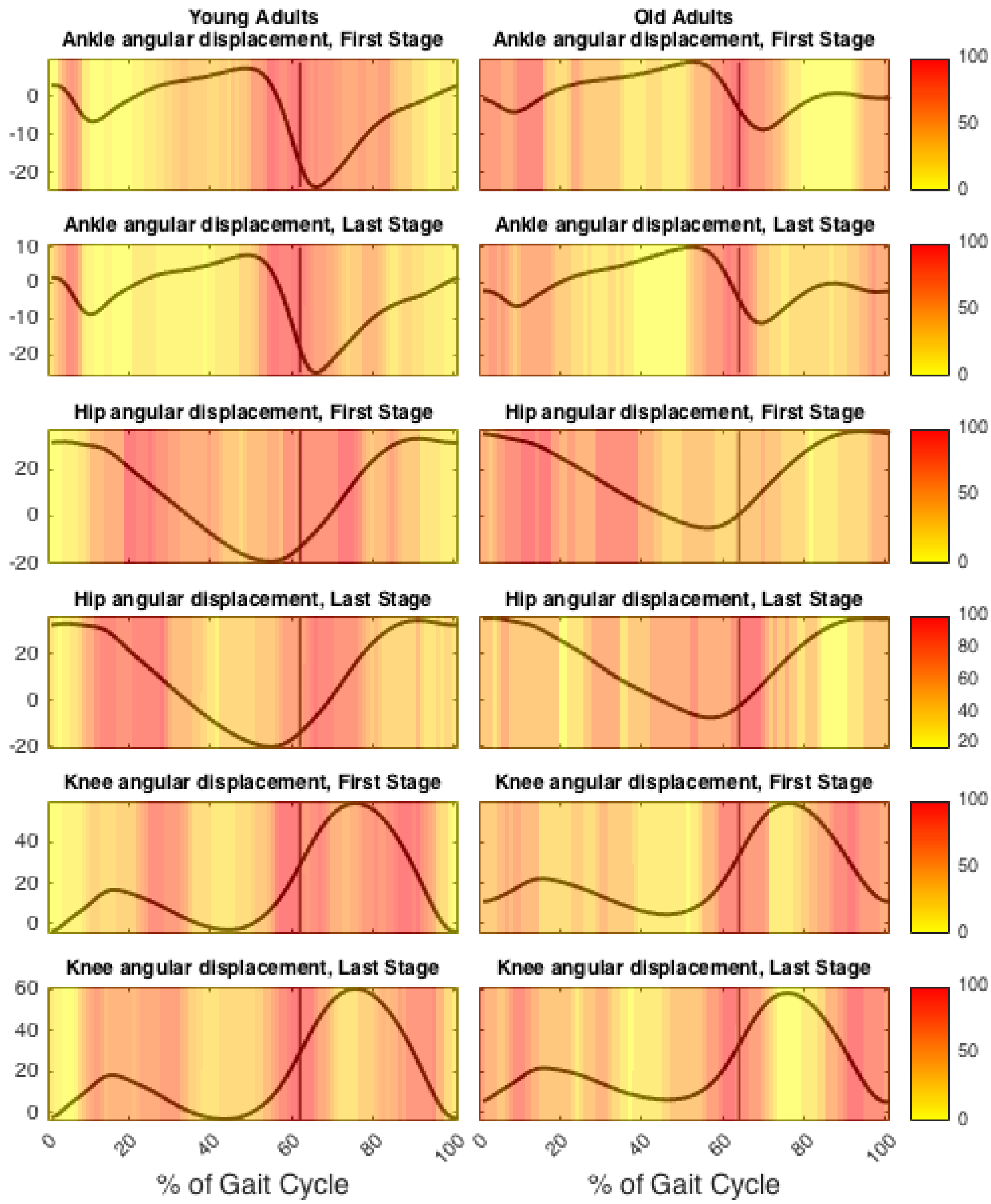

Figure 1: Mean ensemble curves for the sagittal plane angular displacement at the ankle, knee and hip during the gait cycle. The color scale was used to illustrate the intra-cycle region with higher variability among the subjects. The gradient of colors represents the percentage of the sample that had values higher than their respective MeanSD at that particular time-point within the gait-cycle.

profiles of the ankle, knee and hip sagittal angular velocity revealed few changes for both age-groups. The only exception is the overall increment of the intra-cycle variability of the hip angular velocity of the old group. In addition, the region with more variability in the angular velocities profiles seem to be concentrated in the swing-phase (Figure 2).

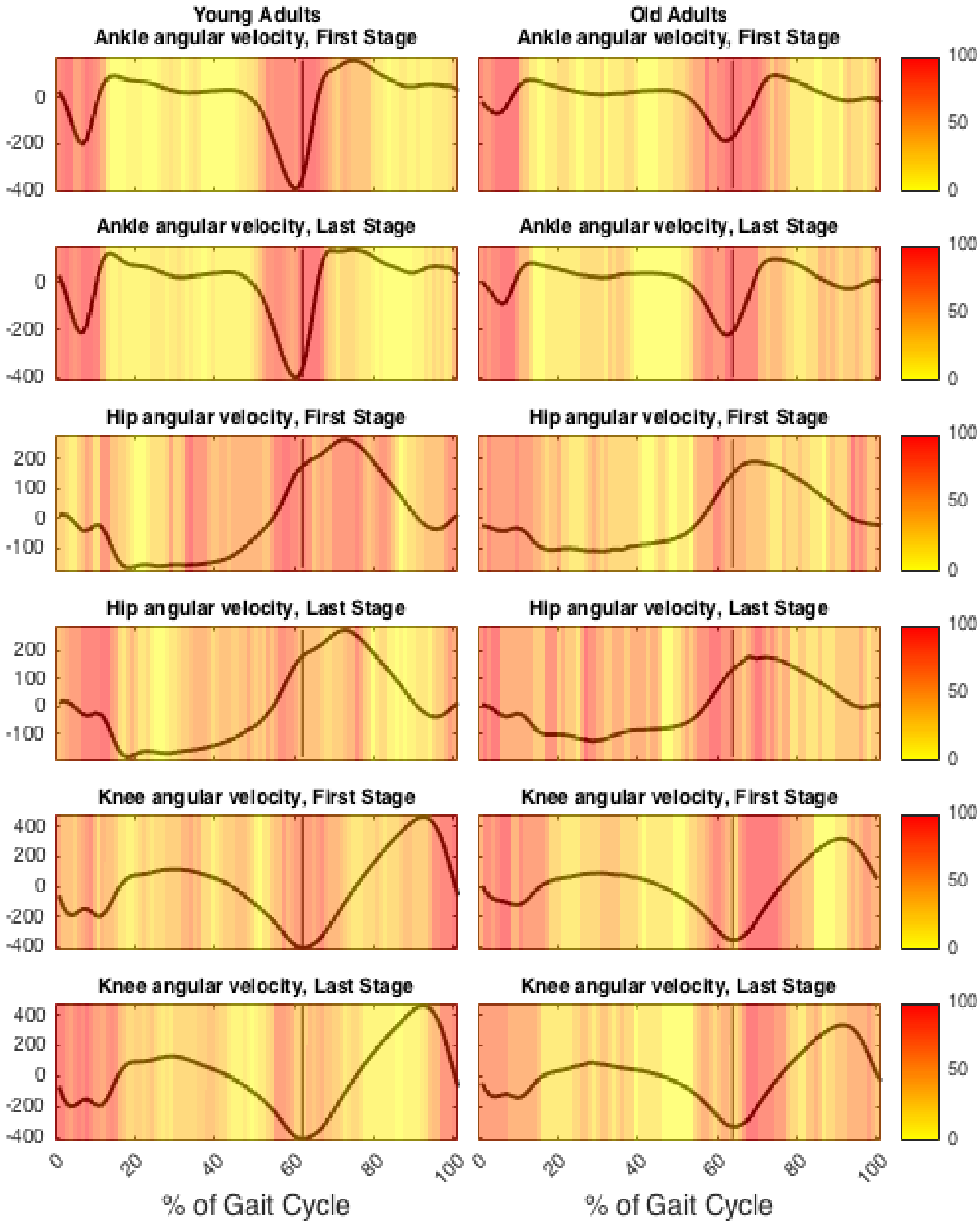

Figure 2: Mean ensemble curves for the sagittal plane angular velocity at the ankle, knee and hip during the gait cycle. The color scale was used to illustrate the intra-cycle region with higher variability among the subjects. The gradient of colors represents the percentage of the sample that had values higher than their respective MeanSD at that particular time-point within the gait-cycle.

The EMG intra-cycle variability, similarly, was higher during the stance phase, with visible reduction during the swing phase, when the end of the test was compared with the first time-point. Old adults, however, seems to demonstrate more variability throughout the entire gait cycle, except for the TA muscle, where the pattern of variability seems to be higher only during swing phase. Overall, some variability can be noticed during the swing phase in the GM, VM, and TA (Figure 3). In general, for both groups it is visible an increment in the variability during mid-swing and towards the end of the swing phase (Figure 3).

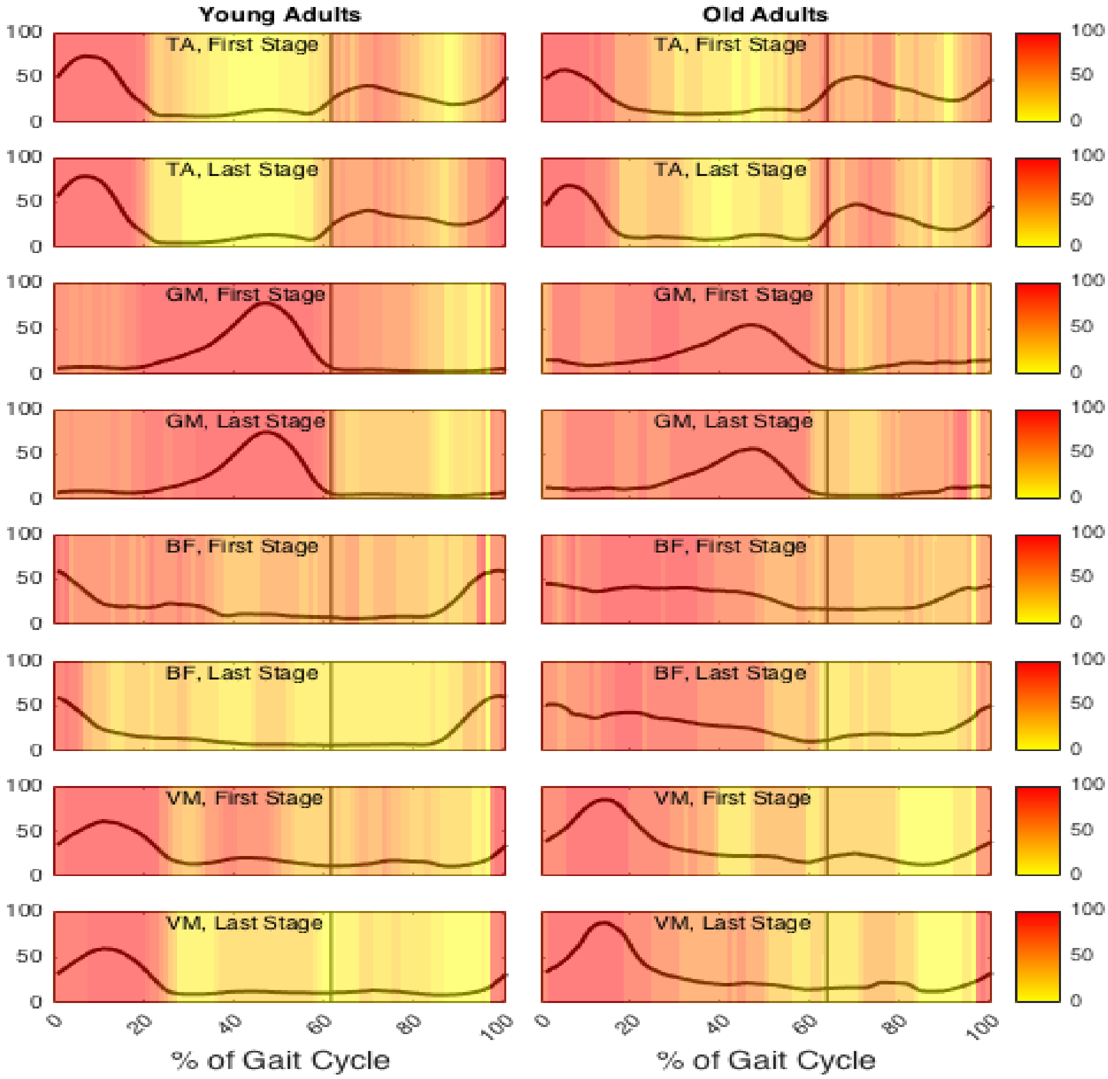

Figure 3: Ensemble averaged electromyographic activities for TA, GM, VM, and BF during the gait cycle. The color scale was used to illustrate the intra-cycle region with higher variability among the subjects. The gradient of colors represents the percentage of the sample that had values higher than their respective MeanSD at that particular time-point within the gait-cycle.

5.4 Discussion

The main purpose of this study was to evaluate the EMG and joint angular kinematics variability along sustained fast-walking activity. For the EMG patterns, it was observed a reduction of the interstride variability of TA and GM along the task. Meanwhile, kinematic variability measures did not show significant fluctuations across time. In the present study, no comparisons were made between variability values before the task (at the baseline). Previous studies demonstrated that the variability of EMG and kinematics increased with speed (Kang & Dingwell, 2008; Kang & Dingwell, 2009; Raffalt et al., 2017). Therefore, we assumed that our subjects started the protocol with higher values of variability between strides compared to freely chosen gait speed, what is in agreement with the theory of dynamic systems which proposes that motor systems self-organize to find the most stable solution for producing a given movement (Hamill et al., 1999). Throughout the protocol the kinematic patterns kept their level of variability, while the calf muscles lowered their variability towards the end of the activity. It may suggest that the hustle of the motor system, to maintain the motor output and keep it within safety boundaries, persisted throughout the activity.

In the present study, variability was calculated considering the average standard deviation over the gait cycle, and then compared across time. This method accounts for the overall variability expressed along the gait cycle and it has been used elsewhere (Hamacher et al., 2017; Kang & Dingwell, 2008; Kang & Dingwell, 2009; Raffalt et al., 2017). The reduction in the variability may result of an effort to achieve the stability of the gait along the task, keeping the motor output performance within a safety pattern (Roos & Dingwell, 2013). Given that a rhythmic movement has greater cycle to cycle reproducibility (Jordan et al., 2007), the reduction in variability along time (or its maintenance) may result from the stability associated with the repeatability of the movement. Fast-walking is highly dependent on neuromuscular function of the calf muscles (Clark et al., 2013). In fact, ankle plantarflexion during stance phase is the largest contributor of energy into the gait cycle (Neptune et al., 2008); so, the reduction in the gastrocnemius variability may traduce an attempt to improve its mechanical performance. Furthermore, the reduction or maintenance of the neuromuscular noise can be an adaptive strategy to reduce risk of falls (Roos & Dingwell, 2013).

One cannot assume that it was not necessary any intra-limb adjustments throughout the protocol because increments in variability of the lower limb joints along the task were not observed. In fact, different values of variability were found in the joint angles, which suggest that the motor output may have more degrees of freedom in one of the components of the kinematic chain. The assumption that the interstride kinematic variability reflects the overall adaptability of the system to the task may not reflect the intra-cyclical demands during the activity. For this purpose, changes in the temporal structure of variability within the gait cycle may provide further information into mechanisms of gait control (Tanimoto et al., 2016). Usually, biomechanical studies focused on portions of the gait cycle based on the characteristics or demands related to those particular gait sub-phases, which can misguide the analysis due to the generalization of signal pattern among the individuals. The characteristic of the variability within the gait cycle may help to address the analysis related to the biomechanical adaptations of interest. In the present study, it was used a modified version of the stride-to-stride variability proposed by Dingwell and Cavanagh (2001) to identify sites with more variability within the gait cycle among the group of subjects. In fact, the kinematics and neuromuscular pattern have shown either high or low variance within the gait cycle. Intra-stride changes have been demonstrated by the literature, and can reveal potential sites of higher variability during the gait (Hamacher et al., 2017). Yet, intra-cycle potential sites of instability in electromyography signals had not been described so far.

The local neuromuscular noise observed herein for both age-groups coincides with a portion of the phase where the muscle is normally active within the gait cycle. Meanwhile, in the beginning of the exercise we can notice some degree of variability in portions of the gait where the muscle is expected to be inactive, which can suggest portions of the gait with increased coactivation. However, along the protocol, this gradient of variability seems to be reduced in both groups. These findings may explain the overall reduction in the coactivation levels due to fast-walking we previously reported. Yet, most of the differences in coactivations levels were noticed during the swing phase, and close to the next heel strike (Oliveira et al., 2017). The kinematic intra-stride variability revealed higher portions of variability towards the beginning of the stance phase. Accordingly, Ihlen et al., (2012) reported that old adults were more unstable during the weight transfer compared to younger persons.

Some studies claim that when the system is exposed to fatiguing tasks there is a re-organization of motor strategies (evidenced by alterations in variability) so that performance can be preserved (Fuller et al., 2011; Skurvydas et al., 2010). Moreover, fast-walking has been proven to induce fatigue, and can be considered a maximal exercise capacity test in healthy subjects (Gonçalves et al., 2015; Nagano et al., 2014). Thus, the findings reported herein caused by the prolonged time walking at a faster pace, may have been caused ultimately by fatigue.

In general, both old and young adults maintained their kinematic variability levels throughout the exercise. With respect to their muscle activation pattern, ankle muscles reduced its variability during the task in old adults. Kinematic and electromyographic intra-stride variability revealed portions of the gait cycle with higher variability. These intra-cycle regions of higher variability seem to reduce along the task, except for higher variability observed during the swing phase and immediately prior to heel strike. Nevertheless, this method only identifies the intra-cyclical regions of higher variability among the subjects. Further studies should quantify the amount of change in gait variability. Some limitations of this study were the small sample size and the use of a treadmill, which may alter the natural variability (Dingwell et al., 2001). The experiment was conducted on a treadmill to collect continuous gait cycles and better control gait speed.

In summary, analysis of interstride variability demonstrated an overall maintenance of the motor output. However, changes in intra-stride variability seems to suggest a shift in the motor strategy during the activity, and additionally, revealed increased variability in portions of the gait cycle associated to risk of falls by tripping or slipping. Future research could include an investigation of the influence of fast-walking on gait parameters related to risk of falling.

Conflict of interest

The authors declare that no conflict of interests is associated with the present study.

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Chapter 6

The influence of fast-walking on the variability of slip and trip-related kinematics

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Abstract

Previous experiments have demonstrated that walking at a faster speed can induce changes on gait that may increase risk of falls. Such type of activity, involving the whole body, adversely affects balance and increases gait variability. The objective of this study was to investigate the effects of fast-walking on the amount of variability of the gait parameters related to the risk of falls by tripping and slipping. Pearson's correlation was used to verify the influence between the metrics derived from the EMG and the kinematics. Our results revealed the presence of regions of higher variability within the gait cycle. The heel horizontal velocity reduced their variability before the heel strike in almost all the subjects. And, the toe vertical displacement reduced their variability towards the time of its minimal height. These alterations suggest that adaptations to a perturbation are phase-dependent, and may indicate a safety mechanism to avoid a fall incident. Changes in the association found between the angular kinematics and the movement control of the toe vertical trajectory and heel horizontal velocity suggest a shift in the motor strategy during the activity.

KEYWORDS: VARIABILITY, FAST-WALKING, TRIPPING, SLIPPING, AND ELDERLY

6.1 Introduction

Falls are the leading cause of injuries in people aged 65 years and above. Since most of falls have been reported to occur during locomotion, the mechanics of gait has received a lot of attention. Previous experiments have demonstrated that walking at a faster speed can induce changes in gait that may increase risk of falls (Nagano et al., 2014). Such type of activity, involving the whole body, adversely affects gait stability (England & Granata, 2007; Kang & Dingwell, 2008).

A certain level of variation is expected in any repetitive movement. In human motion, variability is considered intrinsic to any repetitive motor pattern. This intrinsic movement variability reflects a strategy of the motor control system of performing a motor task within some degrees of freedom (Stergiou & Decker, 2011). And, yet, when exposed to a physical constraint, the system is expected to increase the levels of variability to later lower its variation as it adapts itself to the new order. In human walking, a fall may take place when a perturbation induces an instability in the system beyond the individual's capacity to overcome the perturbation inflicted. Thus, a fall event depends on the individual's neuromuscular capacity, as also on the type and intensity of the perturbations encountered in daily life.

When walking at a faster speed, old adults increases gait and neuromuscular variability (Kang & Dingwell, 2008; Raffalt et al., 2017), and greater joint kinematic variability during the swing phase has been consistently associated with risk of falls (Kobayashi et al., 2014; Mills et al., 2008). Additionally, perturbations on gait can be observed during specific phases of the gait cycle, revealing potential sites of higher instability (Ihlen et al., 2012). In a previous work, we reported that walking at a faster speed induced local alterations in the gait pattern where regions with increased level of variability were noticeable along the gait cycle, more particularly during the swing phase. It is particularly during this phase of the gait that a fall event may take place. At mid-swing, the toe reaches its lowest vertical distance, where the chance of tripping is greater (Winter, 1992). Immediately before the heel touches the ground, high horizontal heel velocity is thought to increase the risk of slipping (Karst et al., 1999). However, the effect of walking at a faster pace on the variability of gait parameters related to the risk of falls by slip or trip is yet to be established.

Previous studies have analyzed variability of a movement pattern through the generalization of the fluctuation values within the gait cycle. The assumption that the interstride kinematic variability reflects the overall adaptability of the system to the task may not reflect the intra-cyclical demands during the activity. For this purpose, changes in the temporal structure of variability within the gait cycle may provide further information into mechanisms of gait control (Tanimoto et al., 2016). The characteristic of the variability within the gait cycle may help to address the analysis related to the biomechanical adaptations of interest. The main objective of the present study was to investigate the effects of fast-walking on the amount of variability of the gait parameters related to the risk of falls by tripping and slipping. For this purpose, we used principal component analysis to investigate intra-individual variability due to fast-walking within the gait cycle. And, additionally, to determine whether the kinematic and neuromuscular variability were reflected in the fluctuations of the kinematics associated with tripping and slipping risk.

6.2 Methods

6.2.1 Subjects

The participants 8 old adults (69.6 ± 3.8 years; 1.56 ± 0.1 m; 66.9 ± 8.2 kg), free of any orthopaedical, neurological, visual, vestibular or cardiovascular conditions that would prevent the subject to perform all the proposed activities. All participants provided informed, written consent. This research was approved by the Local Ethics Committee.

6.2.2 Task and Procedure

Subjects were asked to walk on the treadmill at their self-preferred speed for five minutes to become familiar to the device and walking surface. To ensure the safety during the protocol, the participants wore a safety harness attached to the ceiling. After the familiarization period, the treadmill speed was increased by 0.5 km/h every 30 seconds till the participants reached the speed where they would be at 70% of their age-predicted maximal heart rate (220 minus age, in years). Data acquisition started after a period of one minute at that particular speed. The participants were instructed to walk at this intensity for twenty minutes or until the voluntary exhaustion. Heart rate (HR) was measured continuously during the test with a Polar HR monitor (2010 Polar Electro Oy, FI-90440, Finland) to ensure they were walking at the required intensity. Additionally, the rating of perceived exertion scale was obtained throughout the protocol using a modified Borg 10-point scale (Borg, 1982).

Prior to data acquisition, reflective markers were placed bilaterally over posterior superior iliac spines, anterior superior iliac spines, greater trochanter, medial and lateral femoral condyles, medial and lateral malleoli, posterior heels, medial first metatarsal head. Motion data were recorded using an eight Oqus cameras motion capture system (Qualisys, Inc., Sweden) operating at 100 Hz. Kinematics and EMG were recorded simultaneously. Surface electrodes with an inter-electrode distance of 10 mm, 4-bar formation, and bandwidth of 20–450 Hz, were applied to the midline of the palpated muscle belly, parallel to muscle fiber orientation over vastus medialis (VM), biceps femoris (BF), tibialis anterior (TA) and gastrocnemius medialis (GM), recording sites were previously shaved and cleaned with alcohol, and then the electrodes were affixed to the skin with hypoallergenic tape, according to SENIAM guidelines. The electrodes were further secured to reduce motion artifact using flexible, non-adhesive wrap encircling the thighs (Retelast™ Tubular Stretch Bandage Large, 3M Healthcare). Surface EMG was recorded using a Trigno wireless acquisition system (Delsys Inc., Boston, MA, USA) at a sampling rate of 1000Hz.

Data were exported to Visual 3D (v.5, C-motion, USA). Marker information was filtered using a Butterworth low-pass 4th order filter at a 12 Hz cut-off frequency. Heel strike was defined as the maximum anterior trajectory of the heel marker. Gait cycle was defined by consecutive heel strikes of the right foot. The position of the virtual toe marker was computed during post processing. The virtual toe marker was defined at the anterior edge of the shoe by reconstructing its position relative to the markers placed at the distal aspect of the shoe and the 5th metatarsal head. Ankle, knee and hip joint kinematics were quantified using an XYX sequence of rotations (where X is flexion-extension; Y is ab-adduction and is Z is internal-external rotation). Minimum toe clearance was defined as the minimum vertical height of the virtual toe marker during the swing phase. Toe vertical displacement was obtained from the virtual toe marker of the dominant leg. The horizontal heel velocity was calculated using the heel marker horizontal position first derivative. Heel contact velocity was extracted at 1/100 seconds before the heel strike. Sagittal joint angles at the ankle, knee and hip, its respective angular velocities and EMG were time normalized to 100% of the gait cycle. EMG signals were bandpass filtered (Butterworth digital, fourth order, 20 Hz - 500 Hz), rectified and the RMS of the signal was obtained using 100ms moving continuous windows. All EMG waveforms were amplitude normalized to the maximum EMG amplitude obtained for each muscle during the first 30 seconds of the trial. The last thirty seconds of the activity were compared to the 30 seconds obtained at beginning of the task.

PCA approach

Principal component analysis (PCA) was performed in ankle, knee and hip sagittal angular displacement, respective angular velocities, vertical displacement of the virtual toe marker, heel horizontal velocity and EMG waveforms according to (Soares et al., 2016). In summary, the aim of PCA is to summarize the information contained in 100% of the kinematic curves in a smaller number of components that explain the greater variance through linear combinations from those variables, by considering each 1% in time axis as one variable, and to represent the full waveform by a smaller number of components (PC model - matrix Z) that explain most of the variance through linear combinations from those variables (Jolliffe, 2002). The PC models defined by the equation $\boldsymbol{Z = UtX},$ where U are the eigenvectors of the covariance matrix of X (matrix S). Un is calculated by the equation $\boldsymbol{SUn = \lambda Un}$ where λ are the 100 eigenvalues.

PC model calculation: In this work, the individual's stride-to-stride local variability of lower extremity motion and EMG patterns during fast-walking was examined using PCA. The generated PC model (Matrix Z) is a 100x100 matrix, determined by the product of U (100 eigenvectors) and X (100 columns representing the kinematics for each variable for each subject). From matrix Z, the first three columns (PC1, PC2 and PC3) were retained for analysis, since according to a previous study (Jolliffe, 2002), the first 3 PCs contain the most variability explained. This procedure was performed 12 times, one for each of the analyzed outcomes (i.e. PC model from the (i) ankle, (ii) knee, (iii) hip sagittal displacement, (iv) ankle, (v) knee, (vi) hip angular velocity, (vii) toe vertical displacement, (viii) heel horizontal velocity, muscle activation patterns of the (ix) tibialis anterior, (x) gastrocnemius medialis, (xii) vastus medialis, and (xii) biceps femoris.

PC scores calculation: The PC score values (sn) are obtained by applying the equation sn=ZA, where A is all the matrix (n repetitions for each subject times 100 columns of data, equivalent to matrix X, containing the data from the condition where the model is expected to be applied, (i.e. each subject has its own PC model). This procedure generates a vector (1 line times n repetitions) of data where each curve (each repetition in each variable) is represented by a number (score).

In summary, for the determination of the correlation between variables, the PC model (matrix Z) was developed based on the pattern of the subjects in 30 seconds at the beginning of the protocol. The PC score values (internal product from Matrix Z PC1, PC2 and PC3 to matrix A) for each subject in each condition were retained for analysis (PC1, PC2 and PC3 scores for the 12 waveforms, totalizing 12 PC score values per subject in the first stage and 12 in the last stage). In the last phase, load vectors were calculated by normalizing the PC models (matrix Z) between -1 and 1 according to (Jones et al., 2008). After normalization, a threshold of ± 0.71 was adopted to consider a load vector from one variable as relevant, and then to attribute a meaning for this PC (Knapp & Comrey, 1973). It means that a variable only with values above these loadings, have a biomechanical interpretation in that portion of the curve (Jones et al., 2008).

6.2.3 Statistical analyses

Posteriorly, Pearson’s correlation was used to describe the relatioship between the variability values obtained with the PCA scores derived from the EMG and the kinematic curves. A paired t-test was applied to analyze the differences between mean scores at the beginning and at the end of the activity of the minimum toe clearance (MTC) and heel contact velocity (HCV). All analyses were performed in SPSS (version 24, IBM SPSS, Chicago, IL.), with significance accepted at p <0.05.

6.3 Results

Correlation analyses were conducted to examine the relationship between the toe vertical displacement and horizontal heel velocity variability with various potential predictors from kinematic and muscle activation. Table 1 summarizes the analysis results. Also, the differences in the PC scores of the TVD and HHV were tested to analyze the influence of the duration of the exercise in the internal variability, see figure 1. At the end of the activity, median values of minimum toe clearance significantly reduced (p < 0.023), while mean heel contact velocity significantly increased (p < 0.035).

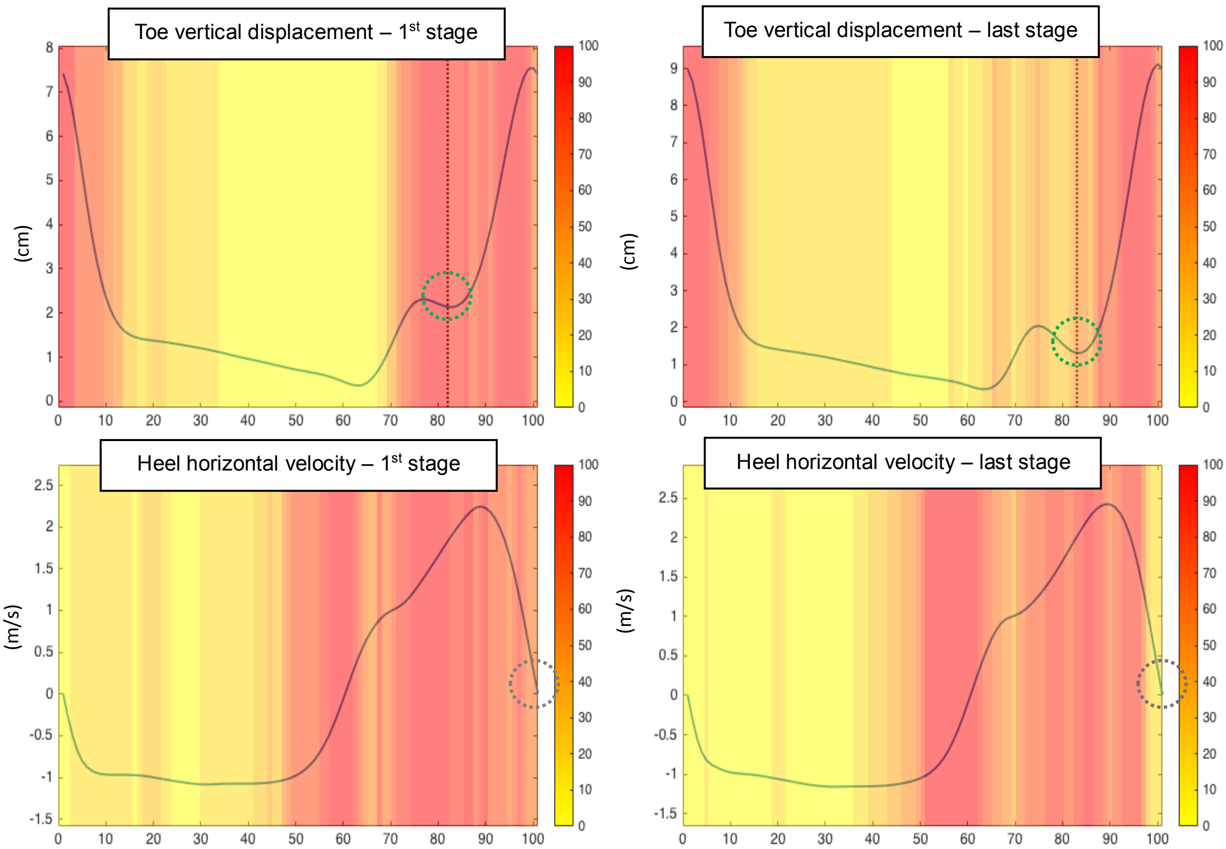

Figure 1: Mean ensemble curves for the heel horizontal velocity, and the toe vertical displacement normalized for the gait cycle (%). The color scale represents the percentage of the sample higher that had values higher than their respective MeanSD at that particular time-point within the gait-cycle. Dashed green circle represents the minimum toe clearance. Dashed grey circle represents the heel contact velocity.

Table 1: Pearson correlation analysis results.

	Heel horizontal Velocity		Toe vertical displacement	
	1st minute of the activity	Last minute of the activity	1st minute of the activity	Last minute of the activity
Ankle angular displacement	.270*	.325**	.142*	.099
Ankle angular velocity	.428*	.518*	.290**	.299*
Knee angular displacement	.180**	.064	.020	.169*
Knee angular velocity	.075	.502**	.076	.159*
Hip angular displacement	.141*	.001	.075	.008
Hip angular velocity	.198**	.241	.258*	-.04
Tibialis anterior	.073	.153*	-.333	.013
Gastrocnemius medialis	.099	.103	.310	.055
Vastus medialis	-.041	.066	.075	-.083
Biceps femoris	.112	.020	.241	.278*
Toe vertical displacement	.379**	.141	-	-

***. Correlation is significant at the 0.01 level (2-tailed).*
**. Correlation is significant at the 0.05 level (2-tailed).*

6.4 Discussion

During fast-walking, perturbations on gait were accounted from levels of kinematic and muscular variability (Jordan et al., 2007; Kang & Dingwell, 2008). In the present study, we aimed to investigate the effects of fast-walking on the variability of gait parameters related to the risk of falls by tripping and slipping. For this purpose, we used principal component analysis to account for the variability within the gait cycle for each participant. Comparison of intra-cycle regions of variability revealed that, at the end of the activity, the majority of the participants reduced the variability of relevant parameters at crucial moments of the gait cycle: (i) the heel horizontal velocity before the heel strike and (ii) the toe vertical displacement towards the time of its minimal height.

In respect of the current findings of local change in variability within the gait cycle, this behavior can indicate the capacity of the system to adapt itself to the new imposed constrains. In the present study, the variability before the fast-walking activity was not measured. Thus, we cannot infer whether the variability found in the beginning of the activity is higher than when walking at preferred walking speed. However, according to the concept of optimal variability in human movements characterized by the inverted U-shape function (Stergiou & Decker, 2011), the reduction in variability observed within the cycle at the end of the protocol can be either a reestablishment of the optimal levels of movement variability, either a robotic behavior due to the tiredness induced by the activity. Considering that reduction in the variability of the minimum toe clearance has been considered a strategy to prevent from tripping (Begg et al., 2007), the local reduction in variability found herein may indicate a safety mechanism to avoid a fall incident. Participants also reduced heel horizontal velocity variability towards the critical period of weight transfer. This behavior, similarly to that of the toe clearance, may also be a mechanism to control the heel movement and prevent a slip initiation.

In this study, we also sought to determine whether the kinematic and neuromuscular variability would reflect a more chaotic pattern in the gait parameters associated with risk of falls by tripping or slipping when walking at a faster speed. For this purpose, we verified which variables were associated with the variability of the heel horizontal velocity and toe vertical displacement. In the present study, the variability of the ankle angular displacement and velocity were consistently reflected on the HHV. The association between ankle angle and the heel horizontal velocity has been reported elsewhere (Lew & Qu, 2014; Moyer et al., 2006). Previous studies have suggested that people may increase ankle plantar flexion as an adaptive postural control strategy to control heel contact velocity as an attempt to reduce the likelihood of slip initiation (Lockhart et al., 2007; Moyer et al., 2006). Given that the tibialis anterior is the main muscle keeping the foot up (dorsiflexed), this safety postural strategy mentioned above could explain the association also found between the variability of the tibialis anterior with the HHV. The moderate association between the knee sagittal velocity and with the HHV were only observed at the end of the activity, which suggest a change in the motor strategy among the lower limb joints to control the foot trajectory.

Similar to the relationships found for the HHV, the variability of the ankle and knee angular velocity also demonstrated to be reflected onto the toe vertical displacement variability. At the end of the activity, the knee angular displacement variability was reflected onto the TVD. Similar findings were reported previously by Mills et al. (2008) for relationships between the ankle and knee angular displacement variability and MTC variability. Noteworthy to mention that, for both HHV and TVD variability, it seems that a shift of influence from the hip to the knee occurred at the end of the protocol. Although, in general, variability at the ankle joint has more influence on controlling the foot, to either its horizontal velocity as its vertical displacement. These findings on HHV and TVD may indicate that different postural control strategies were applied between the end and the beginning of the activity.

It has been previously reported that there are differences in dynamic stability within the gait cycles. Ihlen et al. (2012) described that elderly presents reduced stability towards the weight transfer. Mahmoudian et al. (2016) made similar remarks regarding intra-cycle regions of lower knee stability. Therefore, signals of instability within the gait cycle better indicate the underline motor control adaptation endured by the system. Conventional assessment of gait stability assumes that the perturbation applied to the system is homogeneous throughout the cycle. This assumption is due to the average calculation procedure used on these methods. By adapting the conventional method used to assess gait stability, Ihlen et al. (2012) revealed intra-stride changes within the gait cycle, meaning that the dynamic stability is time-dependent. In consonance with this, previous studies have pointed to changes in the temporal structure of variability within the gait cycle (Hamacher et al., 2017; Tanimoto et al., 2016). However, only a few studies can be found in the literature alluding to the presence of intra-cycle variability. The current study contributes to this body of literature by providing additional information regarding the presence of intra-cyclical variability. However, principal component analysis was the method used to reveal within-subject differences between strides. This technique generates a set of principal components scores which provides an index of significant variance for each time point. This information was posteriorly accounted for all the subjects to reveal regions within the gait cycle with higher variability between the subjects. Thus, in our study, besides revealing the presence of variability within the gait cycle, we have also demonstrated that fast-walking induced changes in these regions of variability within the gait cycle. This suggest that adaptations to a perturbation are phase-dependent. These alterations within the gait cycle may indicate a motor control strategy of prioritizing the consistency of the motor pattern at crucial events of the gait cycle by reducing its variability.

Moreover, when exposed to a perturbation the variability of the task-relevant parameters, the TVD and the HHV, may not reflect the entire information present in motor variability. Therefore, to understand the motor variability of a task-relevant parameter, analyses should be extended to that of the execution variables - components of the movement pattern of interest (Srinivasan & Mathiassen, 2012). In the present research, we took into consideration the above-mentioned reasons when we analyzed the associations between joint angular and neuromuscular variability, the execution components, with the variability of the TVD and HHV, as the end-point movements. And, as we discussed above, between the end and the beginning of the activity, there was a shift of influence among the different components of the movement over the foot trajectory.

There are some considerations that must be addressed when interpreting the results of the current study. First, our subjects were physically active. Therefore, the motor control adaptation found herein may reflect the physical activity status of our participants. Second, reduced variability at these critical events of the swing phase may suggest a safety strategy to prevent accident falls. However, at the end of the activity, analysis of the minimum toe clearance and the heel contact velocity revealed that both parameters reached values considered to increase the risk of falling. Therefore, our subjects seemed to improve the control of the movement. Nonetheless, they may have increased the risk of fall by tripping or slipping.

In summary, our findings suggest that fast-walking caused a shift of motor postural strategy that was reflected on the kinematic pattern of the toe vertical displacement and heel horizontal velocity. The movement variability of these two gait parameters related to the risk of falls showed to be influenced by the movement variability at the ankle followed by the knee. These observations and findings of previous studies suggest that measurements to prevent falls from tripping and slipping should focus on the ankle and knee stability. Yet, additional preventive measures can be addressed by identifying local changes within the gait cycle in response to a perturbation.

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Conflict of interest

None of the authors have any conflicts of interest associated with this study.

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Chapter 7

Older adults make use of different strategies to control toe clearance during fast-walking

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Abstract

The objective of this paper was to verify how the lower-limb joints influence minimum toe clearance (MTC) during fast-walking. It is less clear if old adults adopt safety strategies to prevent tripping during prolonged demanding activity, and which lower-body angles influenced on the toe height. A multilevel regression analysis was chosen to examine the effects of time and lower-limb joint angles upon MTC values in 27 young (age 19 – 33) and 24 old (age 65 – 72) while walking at 70% of their maximum heart rate for 20 minutes. MTC values decreased over time for both groups. The analysis revealed the influence of the ankle angle on minimum toe clearance at the start of the protocol. Along the task, the ankle and the knee angle had significant effects on the toe trajectory of the elderly. Therefore, fast-walking may increase the risk of falls by tripping. In old adults, the reduction in toe clearance values may be related to loss of fine adjustments between the adjacent joints of the ankle and knee.

KEYWORDS: TOE CLEARANCE, TRIPPING, KINEMATICS, WALKING, FALLS IN ELDERLY

7.1 Introduction

Falls are a public health problem and a health threat for old adults (65 years old and more). According to the World Health Organization (WHO, 2015), 30% of people over the age of 65 years, living in community, fall at least once per year, and this proportion increases with age. Most falls occur during walking (Niino et al., 2000), and tripping has been accounting for approximately 50% of all fall's direct cause (Smeesters et al., 2001). Tripping can be defined as an event where the lowest point of the foot makes an unexpected contact with the floor or an obstacle during mid-swing. During the swing phase of the gait the toe position is directly regulated for a combined kinematic chain, consisting of the pelvis, the stance limb and swing thigh, shank and foot segments (Winter, 1992). The swing toe describes its forward trajectory reaching its lowest vertical point, represented by the minimum toe clearance (MTC). This biomechanical event occurs at the highest forward velocity of the swing foot, representing a highly controlled movement to maintain the clearance around a safety value (Winter, 1992).

The minimum toe clearance value ranges between 10-20 mm and are not different between healthy young and old adults (Begg et al., 2007; Mills et al., 2008; Sparrow et al., 2008). Although there are no ageing effects in the minimum toe clearance central tendency, each age group makes use of different biomechanical strategies to keep the toe clearance at similar heights (Mills et al., 2008). These includes reduced step length, increased single/double support time, and reduced range of motions. The literature attributes these gait adaptations to the lower walking speed in the elderly (Levine et al., 2012).

Previous work has examined tripping risk when walking under challenging situations, such as obstacle negotiation (Nagano et al., 2011), uneven surfaces (Menant et al., 2009), and walking under divided attention (Santhiranayagam et al., 2015). The findings revealed increased values of the central tendency of MTC and decrements in variability, as adaptive strategies to ensure the toe height within safety values. However, most of the research examining the association between toe trajectory and tripping risk has been conducted mostly when people were walking at their normal preferred walking speed for short periods of time.

Moreover, even though the toe clearance is a precise-end-point event during swing phase influenced by the joints on both the stance and swing limbs, most studies have neglected how the adjacent structures regulates the trajectory of the swing toe at this precise event, especially when walking under circumstances that increase tripping risk. For these reasons, the purpose of lhe presente study is to describe the alterations of the vertical displacement of the toe at the minimum toe clearance over time during a fast walking activity, and to determine the influence of the swing and stance lower limb joints on altering the toe trajectory.

7.2 Methods

Participants

Twenty-four old adults were recruited from the local community (10 men, mean age 68.1 ± 3.3 years, height 1.67 ± 0.09 m, body mass 77.8 ± 6.6 kg; 14 women, mean age 66.5 ± 3.1 years, height 1.59 ± 0.10 m, body mass 64.9 ± 9.0 kg) and 27 young participants from among the students (11 men, mean age 25.8 ± 6.2 years, height 1.81 ± 0.07 m, body mass 82.6 ± 12.6 kg; 16 women, mean age 27.1 ± 6.1 years, height 1.61 ± 0.07 m, body mass 57.9 ± 6.8 kg). The exclusion criteria were any orthopedic, neurological, visual, vestibular or cardiovascular conditions that would not allow the subject to perform all the proposed activities. All participants gave written informed consent for participation in the study. This research was approved by the Local Ethics Committee.

Task and procedures

All participants completed a familiarization session on an instrumented treadmill (AMTI Inc., MA., USA) walking at their preferred walking speed for five minutes. Older individuals wore a safety harness attached to the ceiling. After a familiarization period of 5 minutes, the treadmill speed was increased by 0.5 km/h every 30 seconds till the participants reached the speed where they would acchieve 70% of their age-predicted maximal heart rate (220 minus age in years). The participants were instructed to walk at this intensity for twenty minutes or until their voluntary exhaustion. Heart rate (HR) was measured continuously during the test with a Polar HR monitor (2010 Polar Electro Oy, FI-90440 Kempele, Finland) to ensure they were walking at the required intensity. A variation of ten percent of target heart rate was allowed. The perceived scale exertion was obtained throughout the protocol using a modified Borg 10-point scale (Borg, 1982).

Kinematic data, including spatial temporal parameters, were collected for 30 seconds every minute until the end of the activity, allowing to posteriorly choose five equally distributed time points for analysis. Time points were coded from 0 to 4 to establish the beginning of the protocol. 3D body segment kinematics data were captured using an 8-camera motion capture system (Qualisys, Inc., Sweden) operating at 100 Hz. Retro-reflective markers were placed bilaterally over posterior superior iliac spines, anterior superior iliac spines, greater trochanter, medial and lateral femoral condyles, medial and lateral malleoli, posterior heels, lateral aspect of the 5th metatarsal head and at the most distal superior aspect of the shoe. Visual 3D (C-motion Inc., Rockville, MD, USA) was used to calculate the kinematic data. The marker trajectories were filtered using a fourth-order low-pass Butterworth filter with a 12 Hz cut-off frequency. Heel strike was defined as the maximum anterior-posterior direction of the heel marker. Toe off was defined as the minimum anterior-posterior direction of the 2nd toe marker (Zeni et al., 2008). Gait cycle was defined by consecutive heel strikes of the right foot. The position of the virtual toe marker was computed during postprocessing. The virtual toe marker was defined at the anterior edge of the shoe by reconstructing its position relative to the markers placed at the distal aspect of the shoe and the 5th metatarsal head. 3D kinematics of the lower-limb angles was calculated using an XYZ Cardan sequence of rotations (where X is flexion-extension; Y is ab-adduction and is Z is internal-external rotation). The analysis of the vertical toe trajectory during swing phase concentrated on the lowest vertical height of the toe during swing phase of the dominant leg. The following parameters were assessed at the time of minimum toe clearance: minimum toe clearance values, frontal angle of the hip and sagittal angles of the ankle, knee and hip of the swing leg, and foot angle from the stance leg were recorded. For the sagittal plane angles, positive values denote hip flexion, knee flexion and ankle dorsiflexion, whilst negative values correspond to hip extension, knee extension, and ankle plantarflexion. Positive values of the frontal hip angle denote hip adduction, whilst negative denote hip abduction. Foot angle establishes the foot position with respect to the floor, negative values represent the forefoot lower than the rearfoot, and positive values represent the rearfoot is lower than the forefoot. The joint angles included in the analysis were set to those that were previously reported as parts of the lower-limb kinematic chain controlling the toe trajectory (Winter, 1992).

Data Analysis

Prior to the multilevel regression analysis, descriptive statistics and the distribution of all dependent variables were computed by the statistical program SPSS (version 21, IBM SPSS Statistics/9.0; IBM SPSS, Chicago, IL). Thus, within-subjects mean and standard deviation were used for all the lower limb angles, and for the MTC data, within-subject medians and interquartile ranges were used as measures of central tendency and variability, respectively. In addition, separate two-way factorial ANOVAs were used to verify angular adaptations over time between groups, and non-parametric analysis (Mann-Whitney) was used to compare walking speed during protocol between age groups.

A multilevel regression analysis was chosen to examine the associations of age group and lower-limb joint angles upon MTC median values, and the analysis was done in SuperMix software (Don Hedeker, 2008). A series of hierarchical nested models were fitted to explain variation in the MTC values, and the deviance statistic was used as a measure of global fit. At each stage of modelling, a chi-square likelihood ratio test is used to assess if the additional variables improve the model. Modelling was done in a stepwise fashion. The variables included in the modeling analysis were: time points, age group (young adults and old adults), lower limb angles, and the interaction between age group, the lower limb angles of the hip, knee ankle, and foot at the moment of MTC, and time. In a first stage, a random intercept and random slope model were fitted to the data to estimate the variance accounted for by the time effect in MTC. In the step two age group, the lower limb angles were added to the model to determine its influence on the values of the MTC in the beginning of the protocol. As previously determined in the literature each age group makes use of different biomechanical strategies to keep the toe clearance at similar heights (Mills et al., 2008). Moreover, it is possible that the trend in MTC scores over time may differ due to angular variations along the protocol within the age group. To evaluate this potential relationship, it was included in the model 3 variables representing the interaction between the time of measurement and the joint angle interaction with the age group. To facilitate the interpretation of these predictors, all but age groups (0 = Young adults; 1 = Old adults) were centered at the grand mean. Statistical significance was set at $p < 5\%$.

7.3 Results

Walking speed during the protocol was significantly greater in the young group (young 1.91 ± 0.15 m/s vs older 1.67 ± 0.21 m/s). Ageing and time effects on lower-limb angles are presented in figure 1. Ankle angle assumed significant differences between groups right practically during the entire task, with significant increments in plantarflexion of the young group. Despite no changes in the knee along the time, towards the end of the protocol knee angle values were significantly different between groups. Great frontal hip angle variability can be observed in both groups, with a punctual moment of significant difference between them. Old adults showed greater hip flexion at the beginning of the exercise, then assumed values similar to the young counterpart, with both group influenced by time. The foot angle of the stance leg of both groups showed a trend for having the heels higher than the forefoot, this trend became more prominent for the young group throughout the protocol (see Figure 1).

Multilevel modeling results are shown in Table 1. The model 1 contains the final estimates of the fixed and random coefficients included in the model. The results show a significant coefficient (β = - 0.5mm; $p < 0.001$) for the time effect. At the beginning of the exercise, when time point was established as zero, the average MTC value for all the subjects is 1.4 cm. For each subsequent time point, a decrease of 0.5 mm in average MTC value is expected. At the end of the protocol, the average expected MTC value was 1.40 – (0.05 x 4) = 1.2 cm. In other words, in average for both groups we can expect 14.2% of reduction in their toe clearance values. The inclusion of explanatory variables in the model 2 proved to fit the data significantly better than model 1 (∆ Deviance = 59.1, with 6 df, $p < 0.05$). The results reveal that no age difference in toe clearance was found in the beginning of the exercise (p = .91). Regarding the relationships between joint angles on the MTC values, the results reveal that at the beginning of the exercise, only the ankle angle have influence on the toe minimum height at the beginning of the protocol (β = 0.2 mm; $p < 0.05$).

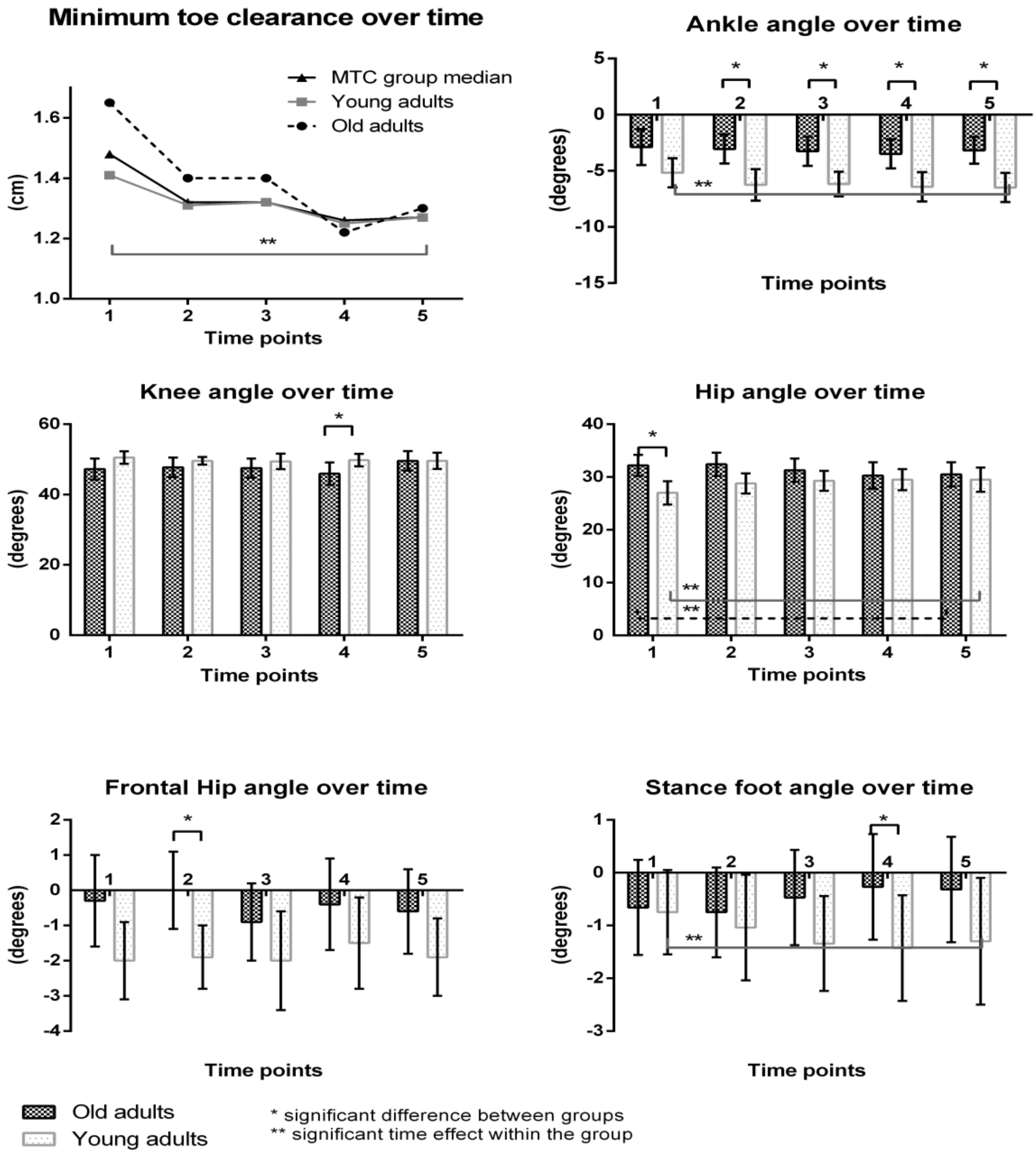

Figure 1: MTC and Lower limb angles between groups over time

The last step intended to investigate the expected change in MTC values at every time point for a unit change in any of the angles of an old adult, by the inclusion in the model of the interaction between time x age group x lower limb joint angles. The MTC values revealed to be sensitive to changes along the protocol in the knee and ankle angles ($p < 0.001$ and $p = 0.01$, respectively). Note that when adding more covariates to the model, the ankle angle no longer has influence on the toe trajectory at the beginning of the protocol ($p = 0.1$). According to the results, the model 3 fits the data better than the model 2 (Δ Deviance = 20.9, with 5 df, $p < 0.05$).

Table 1**.** Summary of results of multilevel modeling for MTC values.

Parameters	*Model 1*	*Model 2*	*Model 3*
	β (SE)	β (SE)	β (SE)
Regression coefficients (Fixed Effects)	1.40 (0.08) $p < 0.001$	1.35 (0.11) $p < 0.001$	1.39 (0.11) $p < 0.001$
Time points	-0.05 (0.02) $p < 0.001$	-0.047 (0.02) $p < 0.01$	- 0.06 (0.01) $p < 0.001$
Age group		-0.02 (0.17) $p = .91$	- 0.1 (0.2) $p = .56$
Ankle angle at MTC		**0.02 (0.01) $p = 0.04$**	0.01 (0.03) $p = .1$
Knee angle at MTC		-0.001 (0.01) $p = .86$	- 0.006 (0.001) $p = 0.37$
Hip angle at MTC		0.002 (0.01) $p = .83$	0.007 (0.01) $p = .35$
Stance Foot angle (SFA) at MTC		0.02 (0.03) $p = 0.16$	0.02 (0.02) $p = 0.3$
Hip frontal angle (HFA) at MTC		0.03 (0.02) $p = 0.18$	0.02 (0.02) $p = .09$
Ankle angle x Age group x time			**0.01 (0.004) $p = .01$**
Knee angle x Age group x time			**- 0.012 (0.01) $p < 0.001$**
Hip angle x Age group x time			0.002 (0.003) $p = .27$
SFA x Age group x time			-0.007 (0.01) $p = .92$
HFA x Age group x time			- 0.006 (0.01) $p = .33$
Variance components (Random effects)			
Random intercept	0.35 ($p < 0.001$)	0.31 ($p < 0.001$)	0.35 ($p < 0.001$)
Random slope	0.01 ($p < 0.001$)	0.01($p < 0.001$)	0.01 ($p < 0.001$
Model Summary			
Deviance statistics	100.54	59.1	35.45
Number of estimated parameters	6	12	17

Coefficients in centimeters

7.4 Discussion

In the present study, we investigated alterations on the minimum toe clearance over time induced by walking at a faster speed. Based on the findings provided in our study, fast walking significantly affected toe clearance over time. Comparing the present findings with previous work is problematic because of differences in the protocol. Nonetheless, in a study about the influence of speed on the toe clearance, Miller et al (2009) reported that increasing walking speed by 0.5m/s would reduce toe clearance by approximately 2.2 mm (Miller et al., 2009). Given that our subjects performed the entire protocol walking faster than their preferred speed, we can, therefore, infer that our subjects may have started the protocol with lower toe clearance values. Note that, in our study, regardless of the baseline effect of the speed on the toe clearance, the velocity is kept the same throughout the protocol. Reasons other than walking speed were responsible for the reduction of 14.2% in the toe height.

Although there has been extensive research on minimum toe clearance during overground walking regarding activities in which tripping risk is elevated, (Miller et al., 2009; Nagano et al., 2014; Santhiranayagam et al., 2015; Sparrow et al., 2008) no study about MTC was carried out during fast walking for such prolonged time, as performed in the present study. In Nagano et al (2014) , the participants were submitted to a 6 minute period of fast walking, but MTC were collected during preferred walking speed. Comparisons of MTC before and after procedure revealed reduction of the MTC similar to our findings. However, the author attributed this alteration to fatigue. Moreover, fast walking has been proven to induce fatigue, and can be considered a maximal exercise capacity test in healthy individuals (Goncalves et al., 2015; Nagano et al., 2014). Indeed, prolonged gait at preferred speed can also induce fatigue in healthy individuals with low physical fitness level (Pereira & Gonçalves, 2011a). Thus, we conclude that the decrement in MTC values observed in our results was caused by the prolonged effort required by the task, which may have induced fatigue. Nevertheless, although tripping risk increased during the task, both groups adopted a biomechanical strategy to maintain toe clearance within safety values (Begg et al., 2007), as we can notice through the MTC variability reduction along the task.

Additionally, we aimed to identify correlates of the lower limb joints on the toe trajectory. For this purpose, we used a multilevel modeling approach given the dependency of data (repeated measures within the subjects). Besides the significant effect of time, as previously mentioned, our regression analysis revealed that at the beginning of the exercise, ageing had no significant effect on MTC values Similar results were found with subjects walking in their preferred speed (Barrett et al., 2010). Yet, at the baseline, participants with more dorsiflexed ankle would have toe clearance values in average 0.2 mm higher. Furthermore, throughout the protocol, alterations in the ankle and the knee in old adults reflect onto MTC values. When walking, towards the time of the minimum toe clearance the swing knee is extending in a forward rotation while the ankle is dorsiflexing in preparation for the heel contact (Moonsabhoy et al., 2009). According to our results, at the time of MTC, if all the other joints remained unchanged an increment of one degree towards knee flexion along the protocol may result in a reduction of approximately 0.5 mm in toe vertical displacement, at the end of the test. Conversely, more dorsiflexed feet increase the toe height, i.e. one degree towards dorsiflexion may correspond an increment of 0.5 mm in MTC values at the end of the protocol. Lower limb influence on toe trajectory has been studied previously (Begg & Sparrow, 2006; Miller et al., 2009; Moosabhoy & Gard, 2006). Although, due to differences in walking condition and subject pools, only the most general comparison between findings can be made. However, from the previous findings, kinematics of the swing foot has been the most consistently reported joint to affect the toe trajectory.

Previous research has reported gait pattern alterations due to local muscle fatigue experimentally induced (Parijat & Lockhart, 2008). Whilst, the present study has examined gait kinematics effects induced by a high demand dynamic activity, such fast walking. In terms of the biomechanical patterns exhibited at the time of the minimum toe clearance, at the beginning of the protocol, the elderly showed increased swing hip flexion in comparison with the young group, consistent with previous findings (Mills et al., 2008; Parijat & Lockhart, 2008). However, except for the reduction in the swing hip flexion throughout the routine, time did not affect any other lower limb joint mean values, in the elderly. Despite the swing hip angle having been the only alteration at the end of the protocol in the elderly kinematic, according to our regression analysis, it seems that the loss of fine adjustments (or coordinated movement) between the adjacent joints of the ankle and knee accounted for the reduction of the toe trajectory.

In summary, the results suggest that fast walking may increase fall risk by tripping. In old adults, the reduction in toe clearance values may be related to loss of coordinated forward rotation between the ankle and the knee joints. Reduced MTC variability has been consistently reported in the literature as a biomechanical strategy adopted to avoid tripping, same response was found in the current study, regardless of the supposed fatigue induced by fast walking. In future research, it would be important to consider the effect of the surface to provide further understanding.

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Conflict of interest

None of the authors have any conflicts of interest associated with this study.

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In general, healthy old adults are more susceptible to fall in outdoors activities, where walking at a faster pace can be a sporadic practice among this population. Therefore, the detection of negative effects of fast-walking on gait performance is important to prevent falls. The effect of speed on elderly's gait mechanics have been studied intensively in the scientific literature. Despite most of the previous studies reporting that fast-walking increases the risk of falling (Callisaya et al., 2011; Faulkner et al., 2009; Nagano et al., 2014; Pavol et al., 1999), some authors (Fan et al., 2016) sustained that fast-walking can be safely performed as a physical activity exercise by healthy old individuals. Besides the contradictory findings with respect to effects of speed on gait, little is known about the ongoing effects of fast-walking activity on gait parameters. This thesis aimed to fill this lacuna in the literature by evaluating gait alterations taking place during this task. To accomplish that, our aims were to: (i) to describe alterations on gait kinematics during fast-walking in young and old adults; (ii) to investigate the influence of fast-walking on muscle activity of lower-limb muscles of young and old adults; (iii) to determine how variability of gait kinematics and muscle activity change during prolonged time walking at faster speed; (iv) to determine whether the variability of EMG and kinematic patterns during fast-walking were reflected in the variability of kinematics associated with risk of slipping and tripping; (v) to verify alterations on slip and trip-related parameters during the activity; (vi) to determine the influence of the swing and stance lower limb joints on altering the swing toe trajectory.

This chapter includes sub-sections regarding the effects of fast-walking activity in different domains of gait biomechanics, namely: spatial and temporal characteristics of gait; gait angular kinematics; and muscular activity. Before we discuss about the main findings of our research, we included a brief consideration regarding differences found in the target walking speed between groups.

Previous work examining the effects of fast-walking speed on gait characteristics was conducted by submitting the participants to a short period of walking at a faster pace, or has used summarized data recorded during the protocol, neglecting the changes that may have occured during the task. Yet, most of studies have established fast-speed as a determined percentage increment of the individual's preferred walking speed. Such procedure may induce different workloads on the participants. Thus, to ensure uniform energy demand among the participants, the target walking-speed required in the present research was established according to the individual's heart rate capacity. The fast-walking intensity was set to 70% of their individual's maximum heart rate (220 – age in years). Once the participants reached the target walking velocity, the walking speed was kept constant throughout the entire protocol. During the protocol, a variation of ten percent of target heart rate was allowed. The exercise would terminate in twenty minutes or until voluntary exhaustion. The participants would be asked to stop the protocol if the exertion reached 90% of their maximum heart rate. Statistical analyses were conducted to identify alterations induced on gait characteristics over the course of the activity.

Age-related differences were expected to occur and were discussed thoroughly, even though, our objective was to investigate whether the activity would offer a potential threat to old adults for increasing the risk of falls in this population. With respect to fast-walking speed, young and old adults achieved the target speed at significantly different velocities (1.9 ±.15 $m.s^{-1}$, and 1.25 ± .28 $m.s^{-1}$, respectively). Such age-related difference in walking speed was expected, since walking requires greater effort in old adults when compared to their younger counterparts (Hortobágyi et al., 2003). Previous studies about age-related differences in gait speed have demonstrated that the differences in gait speed arise from physiological or neuromuscular limitations in old adults (Anderson & Madigan, 2014; McGibbon, 2003).

In the present study, when comparisons between groups were made, we focused on evaluating the differences encountered by young and old adults in response to fast-walking activity. Given that walking speed was kept constant throughout the protocol, reasons other than speed were responsible for the alterations induced by the activity and will be discussed in later sections of this chapter.

1. **Impact of fast-walking on spatial and temporal characteristics of gait**

Taking into consideration the methodology proposed in the present research, we found that spatial and temporal characteristics of gait changed along the course of the activity (Chapter 3). Both age-groups changed their gait strategy by having fewer but longer strides during the task (Kuo, 2001). In a previous study, Ardestani et al. (2016) sought to investigate the influence of various adjustments of cadence and stride length in fast speed walking condition with respect to joint moments. Their findings showed that increasing stride length to walk faster led to remarkable increases in the peaks of hip and knee joint moments. Therefore, the alterations demonstrated by our subjects of increasing stride length could likely increase chances of medial knee osteoarthritis (Miyazaki et al., 2002).

In a previsous study, by Callisaya et al., (2011), the authors required the participants to walk faster over a walkway path, and then posteriorly verified the association between the subject's walking pattern and their prospective history of falls. Their findings revealed that the risk of falls was greater in those participants with smaller steps and faster cadences. Interpreting our results in line with the findings by Callisaya et al. (2011), it seems that the strategy of long strides and lower cadences adopted by our participants during fast-walking may indicate a healthy walking mechanism.

2. **Impact of fast-walking on the variability of spatial and temporal characteristics of gait**

One of the most common features of a movement is its variability. In the literature, classical mathematical methods as standard deviation and coefficient of variation are used to evaluate the magnitude of variability of a motor outcome. The variability of a signal reflects a strategy of the motor control system of executing a motor task within some degrees of freedom (Stergiou & Decker, 2011). Under normal conditions, the values of these fluctuations are relatively small, reflecting consistency and stability within the locomotor system (Almarwani et al., 2016). However, increased variability of spatial and temporal gait parameters has been shown to predict fall risk in old adults (Hamacher et al., 2011; Hausdorff et al., 1997; Hausdorff et al., 2001).

With respect to effects of fast-walking on variability of spatial temporal parameters, our findings showed that old adults demonstrated higher stride length and stride time variability when compared to the younger counterpart (Chapter 3). Literature regarding age-related differences in gait variability (defined here as fluctuations in spatial temporal gait characteristics) and its relationship with walking speed are contradictory. In a previous study by Kang and Dingwell (2008), the authors investigated whether greater variability in healthy old adults could be attributed directly to slower walking speed. In their findings gait variability in old adults was not affected by changes in walking speed more than that of the young adults. However, old adults exhibited greater variability than the young adults, independently of changes in speed. The authors concluded that the greater variability observed in old adults was more likely to result from loss of strength and flexibility than from their slower speeds. Their findings are in agreement with that of previous studies that suggest that age-related changes in variability rather than be a manifestation of walking speed is more likely to reflect an underlying impairment of the motor system.

Effect of fast-walking over time showed to increase step width variability in both age-groups. Increased step-variability has been associated with risk of falls (Maki, 1997; Toebes et al., 2012), loss of balance (Brach et al., 2005) and lowered postural control (Bauby & Kuo, 2000). Increasing step-variability has been suggested to be a necessary adaptation to maintain lateral stability. The increased step-variability found herein, as effect of fast-walking, may be an early sign of lateral instability. The control of lateral foot displacement is regulated by somatosensory, visual, and vestibular input (Bauby & Kuo, 2000). Such increment in step-width variability may be explained by the reduction of sensory information induced by the activity (Paillard, 2012), which is congruent with the negative effects found on postural control and balance induced by similar walking protocol of previous studies (Derave et al., 2002; Ghasemi et al., 2016; Stemplewski et al., 2012). The negative effect of the activity on the sensory motor system may be explained by the momentary proprioception dysfunction caused by the repetitive eccentric and concentric contractions (Paillard, 2012), and the transient vestibular impairment caused by the constant vertical movements (Fitzpatrick & McCloskey, 1994).

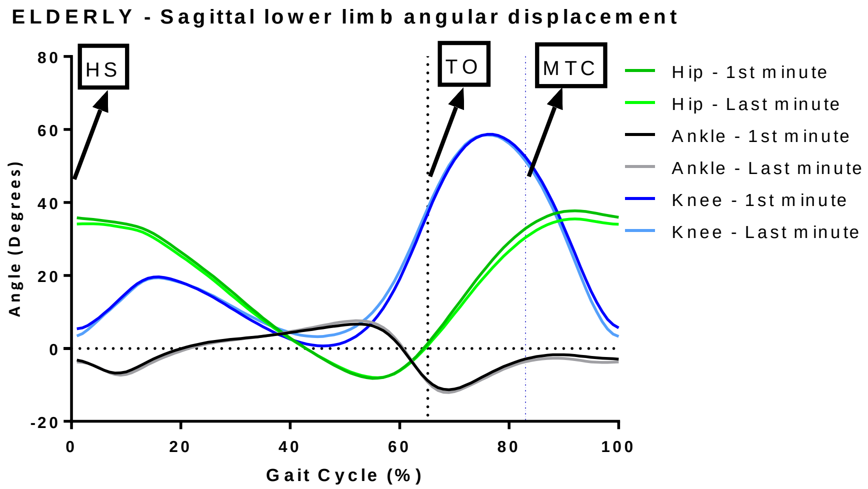
3. Impact of fast-walking activity on lower-limb angular kinematics

In this section, the effects of fast-walking on angular kinematics are discussed separately accordingly to the aspects we have addressed our research question.

3.1 The effect of the activity with respect to lower limb angle position at discrete points of the gait cycle

Although the magnitude of changes in the mean values of the sagittal angular displacement was small for all the joints in both groups, there were significant differences detected by our analysis (see Chapter 3, figure 2). Towards the end of the practice, old adults were initiating the swing phase with additional ankle plantarflexion, and less extended hip (Figure 1). This reduction in hip extension may imply a functional tightness that would be preventing the hip to fully extend at the toe off (Lee et al., 1997). Meanwhile, the increased ankle plantarflexion may be a compensatory mechanism to the reduction in hip extension since the old participants managed to keep and even increase the stride length throughout the protocol (JudgeRoy et al., 1996; Monaco et al., 2009). These findings suggest that, along with the hip flexors, the propulsion power generation progressively combined with the collaboration of the ankle plantar-flexors, which in turn, may be a manifestation of a redistribution of joint torques and their relative contributions to the total motor performance. No changes were observed at the knee at the initial contact and at toe off during fast-walking activity. Thus, the necessary adjustments that may have taken place for weight acceptance and body propulsion were restricted to the hip and ankle joints in old adults.

Our findings suggest that young adults manifested early signs of muscle fatigue, as we could observe through their progressive reduction of the ankle dorsiflexion at heel strike (Parijat & Lockhart, 2008). No other difference was found for the sagittal angles analyzed at the initial contact and at toe off for the young group. Seemingly, the alterations induced by the activity on gait mechanics of the young group were restricted to the ankle joint at landing.



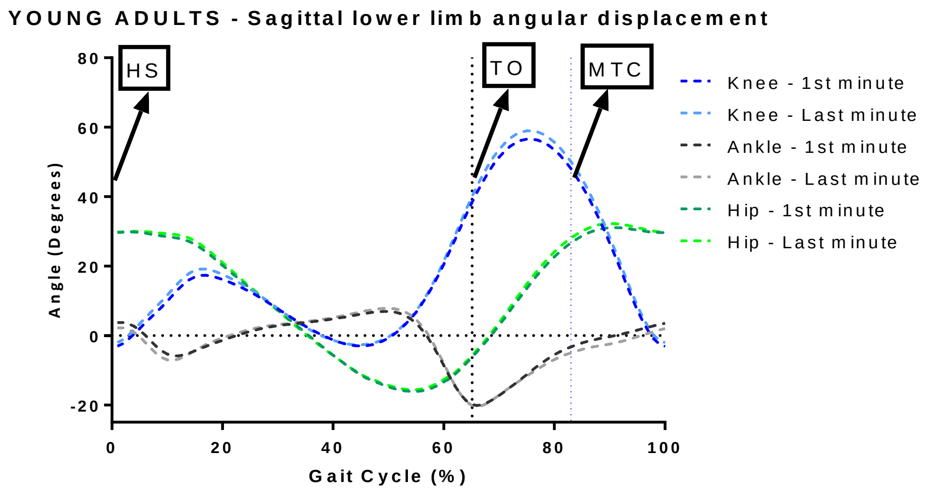


Figure 1: Sagittal angle displacement for the hip, knee, ankle within a gait cycle. Arrows are illustrating the three discrete events at which sagittal angle position were analyzed. HS = heel strike; TO = toe off; MTC = minimum toe clearance.

Given the influence of the lower-limb joint angles on the swing toe trajectory, lower-limb joint kinematics were also evaluated at the minimum toe clearance due to the importance of this event to risk of falls by tripping. In addition to the sagittal angles of the hip, knee, and ankle of the swing limb, we included analysis of the frontal plane of the hip angle for the swing limb and the foot angle with the floor of the stance limb (Chapter 5).

In terms of the biomechanic pattern exhibited at the time of the minimum toe clearance, except for the reduction in the swing hip flexion throughout the task, time did not affect any other lower-limb joint mean values, in the older participants. Meanwhile, the young adults also demonstrated reduction in the swing hip flexion towards the end of the activity. In addition, during fast-walking, the young participant's stance foot slightly increased its vertical distance from the floor, as it can be noticed by the increased in the stance foot angle. Congruent with the trend to drop the foot observed at the heel strike, young adults also increased the ankle plantarflexion at the MTC. Seemingly, the early sign of fatigue persisted from the heel strike to the mid-swing in the young group. The influence of the lower-limb angles on the minimum toe clearance will be discussed separately in the section regarding potential risk of slipping and tripping during fast-walking.

3.2 Changes in intra-limb joint coordination

To better understand the underlying adaptations of the motor control process encountered during the activity, the analysis of a single joint may not be sufficient. The intra-limb coordination pattern can display the coordinative synergism between the adjacent joints during a stride (Shafizadeh et al., 2013). The coordinative pattern is a graphical display of the relative position between the joints in the same plane. In this case, the sagittal plane, since it is in this plane that occurs most of the movement responsible for the forward progression of the body. The intra-limb coordination pattern reflects the organization in terms of time and position of adjacent structures within a stride. In case of an impairment in any component of the motor system, changes in the state of coordination may emerge. Therefore, the graphical display of the intra-limb coordination provides insights regarding potential abnormalities on the gait (Shafizadeh et al., 2013; Verrel et al., 2013). The differences from the 1st to the 5th (last) intra-limb coordination trajectories reflect the changes of the angular kinematics observed during the activity discussed above (and reported in Chapter 3), see figure 2.

For both age-groups, the differences between the trajectories were more accentuated during the swing phase. Therefore, we subsequently carried out analysis regarding the effects of the activity on the angular kinematics at the time of the minimum toe clearance (Chapter 7). Then, recollecting the findings from this analysis, we reported significant differences in the angular kinematics in both age-groups, which explained the shift between the trajectories of the intra-limb coordination during the swing phase.

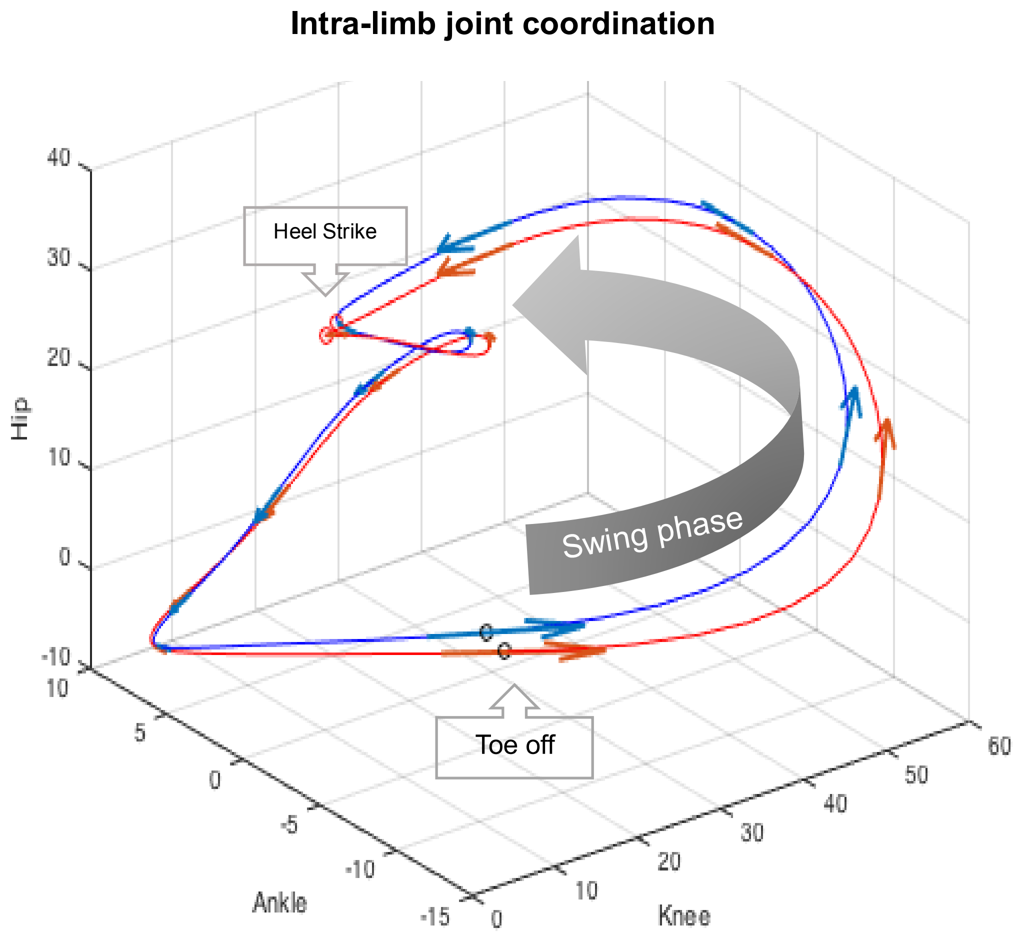


Figure 2: Illustration of Hip angle - Ankle angle - Knee angle intra-limb coordination patterns for the elderly subjects. Blue line represents the mean values of the first stage, red line represents the mean values of the 5th (last stage). These two stages are displayed to highlight the differences induced by the activity. Small arrows indicate the direction of the gait.

3.3 Changes in kinematic variability during fast-walking

In Chapter 3, we presented analysis of the angular kinematics variability at discrete time-points within the gait cycle, namely at heel strike and toe-off. Assessment of variability from the lower limb angular position (assessed by standard deviation) showed old adults with higher variability at heel strike for the ankle and knee than the younger counterparts. Along the protocol, there were no observed differences for the joint angular variability measured at discrete time points.

Then, in Chapter 5, we extended our analysis regarding gait variability to quantify the magnitude of stride fluctuations of the hip, knee, and ankle angular displacement, and their respective angular velocities during fast-walking. Whereby we have quantified variability of the movement pattern by the averaged standard deviation over the gait cycle (100 time-points in a stride). In other words, a global measure of interstride variability within a window frame, in this case a stride. Thus, for each subject the mean standard deviation (MeanSD) characterized the variability for the entire gait cycle of each parameter of interest. Our findings showed that, throughout the protocol, the kinematic patterns kept their variability measures. Different values of variability were found for the joint angles, which suggest that the motor output may have more degrees of freedom in one of the components of the kinematic chain (see Chapter 5).

In the same study, we adapted the analysis of the stride-to-stride variability originally proposed by Dingwell and Cavanagh (2001) to identify changes in the temporal structure of variability within the gait cycle (see methods in Chapter 5). By extending the analysis of variability assessed by the MeanSD, we identified regions of higher variability intra-cycle (see Figure 3) for each subject. Then, by overlapping all the individual graphical display of regions within the gait cycle with higher variability (Figure 4A), we obtained a final graph with a gradient color displaying the intra-cycle regions with higher variability for all the subjects (Figure 4B). To compare the intra-stride variability during fast-walking we displayed graphically the region with higher variability within the gait cycle for all the subjects. The color bar represented the percentage of the total sample with higher variability for that specific percentage of the gait cycle, see figure 4 and 5. This allowed us to identify sites with higher variability within the gait cycle among the group of subjects. Moreover, we observed that during the activity these intra-cycle regions of higher variability seemed to shift within the gait cycle during fast-walking. Overall, the participants reduced the variability during stance phase, although they either kept or increased the level of variability during the swing phase. These findings motivated a subsequent analysis regarding risks of falling by tripping or slipping (Chapter 6), given that these two fall mechanisms are likely to happen during swing phase, more specifically at mid-swing and immediately before landing.

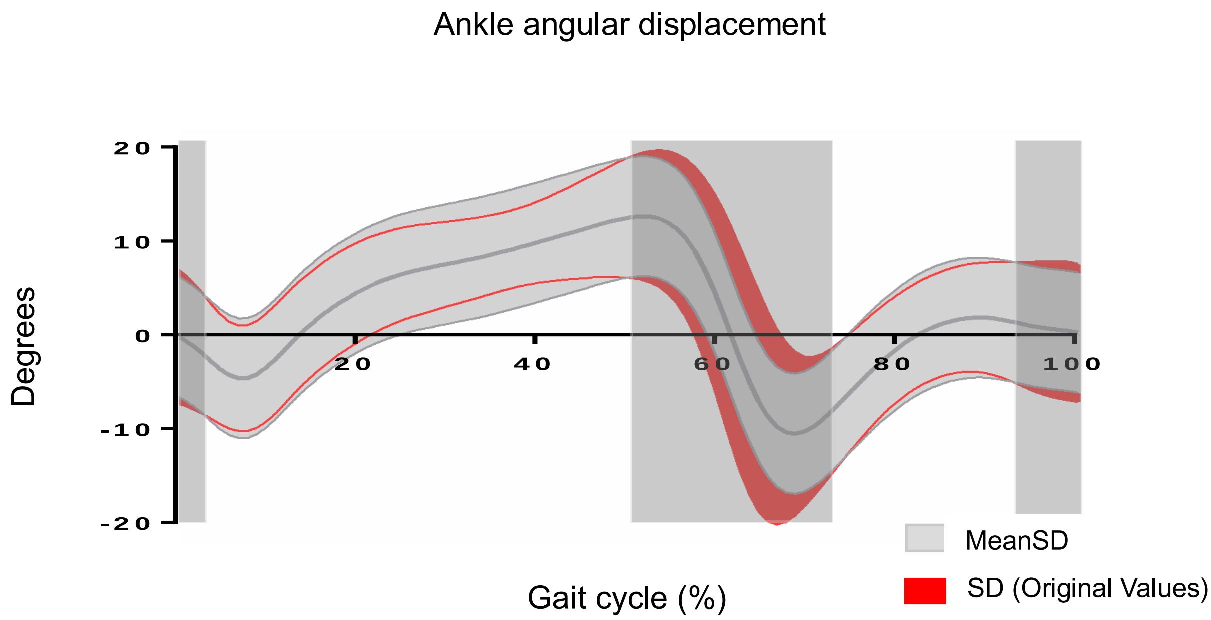


Figure 3: Mean ensemble curve for the ankle angular displacement of an individual. The grey cloud around the mean line represents the standard deviation in its averaged value (MeanSD). The red cloud represents the original values of standard deviation for every percentage of the gait. The region where the orginal standard deviation is higher than the MeanSD was considered regions with a higher variability.

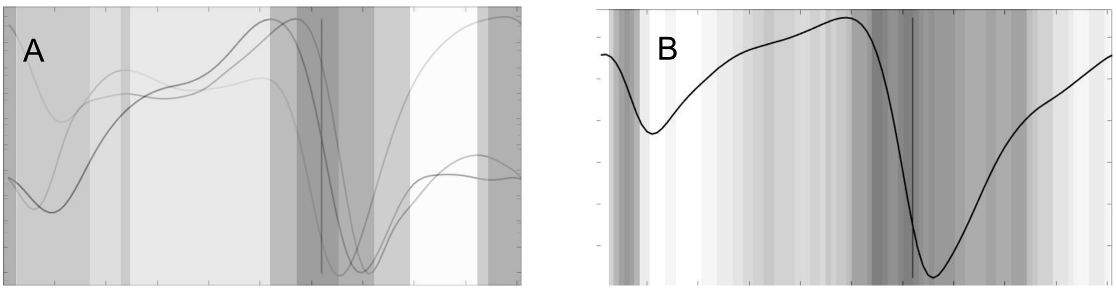


Figure 4: By overlapping the individuals' plots (A), we obtained a gradient color indicating the percentage of the sample with higher variability for each *i*th% of the gait cycle. This information is then displayed over the mean ensemble curve for the variable analyzed (B). The vertical line indicates the average time of the toe off for all the subjects.

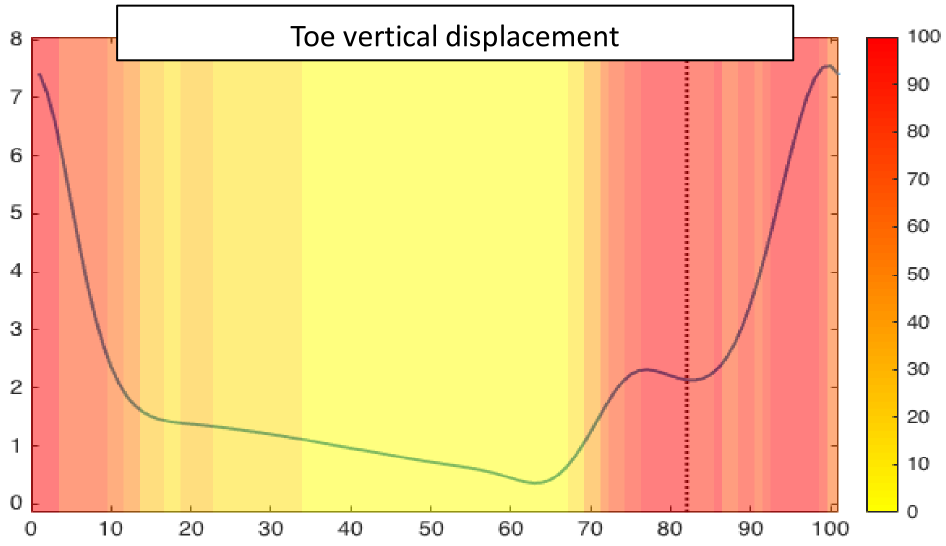


Figure 5: Mean ensemble curves for the toe vertical displacement normalized for the gait cycle (%). The gradient area represents the portion of the gait cycle were the variability is higher among the subjects. Color scale represents the percentage of the sample with values with great variability.

4. Effects of fast-walking on slip and trip-related kinematics

The first studies regarding effects of fast-walking activity on gait have repeatedly pointed to alterations during the swing phase. Therefore, we sought to investigate the effects of the activity on parameters related to the risk of falling by tripping and slipping. In Chapter 6, we investigated the alterations in two aspects of the swing limb foot due to their association to risk of tripping and slipping: the vertical displacement of the swing toe, and the heel horizontal velocity. We used principal component analysis (PCA) to identify changes in the temporal structure of variability within the gait cycle to evaluate perturbations caused by the activity on these two movement patterns.

The PCA revealed regions with higher intra-cycle variability for the toe vertical displacement (TVD) and the heel horizontal velocity (HHV). Comparisons of intra-cycle regions of variability revealed that at the end of the activity the participants reduced their variability at crucial moments of the gait cycle. Interestingly, the HHV, before the heel strike, reduced its level of variability in almost all the subjects. Similarly, the TVD reduced its variability towards the time of its minimal height. Despite the reduction on levels of variability, the minimum toe clearance and the heel contact velocity reached values suggestive of higher risk of tripping and slipping, respectively. Given the reduction of the median values of MTC and the increased horizontal velocity of the heel immediately before landing, the strategy adopted by the subjects to avoid a fall incident was a reduction in the levels of variability, as an attempt to control the foot trajectory (Begg et al., 2007).

Given that the foot trajectory depends on the configuration of the swing leg joint angles, we additionally analyzed the influence of the variability of the swing leg kinematics on the fluctuations of the TVD and HHV. For this purpose, we verified which variables were associated with the variability of the heel horizontal velocity and toe vertical displacement. In the present study, the variability of the ankle angular displacement and the ankle angular velocity were consistently reflected on the HHV. The association between the ankle movement and the heel horizontal velocity has been reported elsewhere (Lew & Qu, 2014; Moyer et al., 2006). Previous studies have suggested that people may increase ankle plantar flexion as an adaptive postural control strategy to control heel contact velocity as an attempt to reduce the likelihood of slip initiation (Lockhart et al., 2007; Moyer et al., 2006). Similar to the relationships found for the HHV, the variability of the ankle and hip angular velocity also demonstrated to be reflected onto the TVD variability. As at the end of the activity, knee kinematics variability was reflected onto the TVD. Similar findings were reported previously by Mills et al. (2008) for relationships between the ankle and knee angular displacement variability and MTC variability.

Of note was, for both HHV and TVD variability, it seems there was a shift of influence from the hip to the knee at the end of the protocol. Although, in general, variability at the ankle joint has more influence on controlling the foot, to either its horizontal velocity as its vertical displacement. These findings on HHV and TVD may indicate that different postural control strategies were applied between the end and the beginning of the activity. In this study, measures of stride-to-stride variability within the gait cycle were found for each subject, and posteriorly used in the correlation analysis with the variables of interest. Thus, in Chapter 6, we showed that kinematic variability was associated with a more chaotic pattern in the gait parameters associated with risk of falls by tripping or slipping when walking at a faster speed.

Moreover, in Chapter 7, we extended the investigation regarding the relationship between the lower limb joints and the minimum toe clearance. In this study, we performed an analysis that considered the correlations between the repeated observations within the same subject. Based on our findings, fast-walking activity was responsible for the reduction of 14.2% of the minimum toe height. At the time of MTC, if all the other joints remained unchanged, an increment of one degree towards knee flexion along the protocol may result in a reduction of approximately 0.5 mm in toe vertical displacement, at the end of the test. Conversely, more dorsiflexed feet increase the toe height, i.e. one degree towards dorsiflexion may correspond an increment of 0.5 mm in MTC values at the end of the protocol. By observing the alterations on the kinematics at the time of MTC over time, we could erroneously associate the reduction on the MTC with the reduction of the hip flexion. However, given that the multilevel regression analysis considers the repeated data at the individual level, this assumption cannot be made. In accordance with previous findings, kinematics of the ankle has been the most consistent reported joint to affect the swing toe trajectory (Begg & Sparrow, 2006; Miller et al., 2009; Moosabhoy & Gard, 2006).

5. Effects of fast-walking on muscular activation and coactivation

Little is known about the neuromuscular adaptations during fast-walking activity. In Chapter 4, we sought to investigate the effects of the exercise on lower extremity muscle activation and coactivation in young and old adults. Comparisons between groups showed significant differences for the tibialis anterior (TA) during the absorptive phase, and for the biceps femoris (BF) during the propulsive phase. Although not statistically different, in general the old adults showed higher levels of muscular activity throughout the exercise. Changes over time were only found for the TA, which increased its values during the absorptive phase and reduced it during the propulsive phase. The reduction of the TA activity along with the slightly increment of the gastrocnemius medialis (GM) during the propulsive phase may explain the increased plantarflexion at toe off found in the Chapter 3. Yet, during walking, the stride frequency strategy is set to minimizes muscular activity (Russell & Apatoczky, 2016). In other words, the reduction observed in TA activity could be associated with the alteration in the gait strategy adopted by the participants (Chapter 3), which in turn is adopted to minimize the overall metabolic cost (Holt et al., 1995).

Overall, old adults exhibited higher coactivation than young adults. Which is consistent with previous studies regarding age-related differences in walking speed effects in coactivation (Hortobágyi et al., 2009; Peterson & Martin, 2010). Age-related differences were attributable to decrements in strength and neurological function in old adults, which may cause this population to increase coactivation to stabilize and stiffen joints in preparation and immediately following weight acceptance. Additionally, at the beginning of the protocol, coactivation indexes for the thigh showed similar distribution across the gait cycle as demonstrated by Peterson and Martin (2010). We found some differences between old and young adults in the distribution across the stride of the coactivation about the shank, which suggest that, at the beginning of the protocol, our old participants relied on the coactivation of the calf muscles to stabilize the ankle during the swing phase. Then, we observed a reduction in ankle coactivation in the first half of the swing phase, to which we have previously suggested to be an attempt to increment the dorsiflexor activity to ensure safety values of the toe height, which proved to be the case, as seen by the reduction in levels of intra-cycle variability of the toe vertical displacement during the respective phase of the gait cycle (reported in Chapter 6).

The reduction of the coactivation levels at the knee during terminal swing could explain the reduction of intra-stride levels of the heel horizontal velocity towards the time of the contact with the ground (as can be seen in Chapter 6). In addition, the increased levels of coactivation levels between vastus medialis (VM) and biceps femoris (BF) during the first half of the swing phase may explain the reduced hip flexion found at the time of the MTC (as reported in Chapter 7).

Posteriorly, in Chapter 5, we found reductions of the interstride variability of the TA and GM during the task. The reduction in the variability may be a result of an effort to achieve the stability of the gait along the task, keeping the motor output performance within a safety pattern, which agrees with what we have previously suggested above when discussing the findings regarding reductions in coactivation levels between the TA and GM. Since the ankle plantarflexion during stance phase is the largest contributor of energy into the gait cycle (Neptune et al., 2008) the reduction in the GM variability may traduce an attempt to improve its mechanical performance.

As discussed in the section regarding kinematic findings above, changes within the gait cycle seems to provide further information regarding adaptations imposed by fast-walking activity. During the activity, the gradient of variability (that express regions of higher variability among the subjects, see further explanation above, in the section 3.3) reduced in both groups. These findings may be associated with the reduction in the coactivation levels during the swing phase discussed previously. Intra-stride changes have been demonstrated by the literature, and can reveal potential sites of higher instability during the gait (Ihlen, Goihl, et al., 2012; Ihlen, Sletvold, et al., 2012). However, intra-cycle potential sites of instability in electromyography signals had not been described before. Thus, we assume that changes in muscle activation variability within the gait cycle can be interpreted according the dynamic theory used for interpreting variability of the kinematics.

Variability of EMG patterns was posteriorly obtained through principal component analysis. By using PCA, we found values correspondent to the variability of the muscle activation pattern for each individual. Thus, after accounting the variability for the total sample, correlation analyses were used to identify associations between the gait parameters of interest and the EMG patterns. Then, it was found that the TA variability slightly influenced heel horizontal velocity, but only at the end of the task. Similarly, at the end of the protocol, variability of the BF marginally influenced the toe vertical displacement. Since ankle kinematics showed to increase its influence on foot trajectory over time (Chapter 5 and 6), the association found between the variability of the tibialis anterior with the HHV may be explained by its muscular function of keeping the foot up. The same rationale can be used to explain the association between BF and toe vertical displacement, since a reduction in BF variability may indicate an attempt to control knee movements.

6. Brief considerations regarding motor variability

The concept of gait stability comes from the presence of an organized repeatability of the level of fluctuation in a physiological signal over time. This organized variability represents the underlying physiologic capability of a healthy biological system to make flexible adaptations to everyday stresses. The variability represents the degrees of freedom of the system in performing a movement. Increments in variability of a signal reflects a strategy of the motor control to adapt to a perturbation inflicted to the system. Methods to quantify stability provide information regarding the variability present in the motor task, or yet, quantify the response of the system to the perturbation inflicted. On the other hand, methods of variability reflect the noise demonstrated by the system, but also the fluctuation in response to a perturbation. Literature frequently refers to this increased variability as the adaptability of the system to compensate for constrains in order to maintain the motor output. Motor variability addresses the variability at different levels of movement execution. Thus, motor variability can be assessed for several types of variables, as task-relevant parameters (e.g. minimum toe clearance); components of the movement pattern like joint angles; muscle activation pattern; and coordinative aspects like the coordination between adjacent angles (Davids et al., 2003; Srinivasan & Mathiassen, 2012; Stergiou & Decker, 2011).

The extent of variability of a task-relevant parameter must be within safety boundaries, otherwise, it may bring the dynamic state of the individual closer to their limits of stability, when the performance of this task is no longer functional. In this context, motor variability analyses can be considered an indirect assessment of gait stability (Hamacher et al., 2011). Moreover, when exposed to a perturbation, the variability of the task-relevant parameters may not reflect the entire information present in motor variability. Therefore, to understand the motor variability of a task-relevant parameter, analyses should be extended to that of the execution variables (components of the movement pattern of interest) (Srinivasan & Mathiassen, 2012). In the present research, we took into consideration the above-mentioned reasons when we analyzed the variability of gait parameters related to the risk of falls (Chapter 6).

In extension to that, previous studies have pointed to changes in the temporal structure of variability within the gait cycle (Hamacher et al., 2017; Tanimoto et al., 2016). However, only a few studies can be found in the literature alluding to intra-cycle variability. The present research contributes to the body of literature by providing additional information regarding the presence of intra-cycle variability (Chapter 5 and 6). We have also revealed that changes occurred in these regions of variability within the gait cycle, suggesting that adaptations to a perturbation are phase-dependent. It seems that the motor control strategy prioritizes the maintenance (or recovery) of stability at specific points of the gait cycle, as we could observe in the study regarding gait parameters related to risk of falls, in Chapter 6.

In summary, during fast-walking walking pattern changed over time for both elderly, and young-group. Although the old adults progressively initiated the swing phase with a more flexed hip, at the time of MTC (mid-swing) the participants reduced hip flexion. With the exception of the significantly increased plantarflexion at toe off, no other differences were found for the old adults over time. Regarding temporal changes of variability within the gait cycle, in general, old adults reduced variability during the stance phase, either keeping or increasing levels of intra-cycle variability during the swing phase. Analyses of variability of task-relevant variables (i.e. toe vertical displacement, and heel horizontal velocity) revealed that at important moments of the gait cycle (at the time of the minimum toe clearance, and immediately before landing) the participants managed to reduce levels of variability. Which seems to be a safety compensatory mechanism for the reduction of the minimum toe clearance and the increment on the heel contact velocity. Given the association found between the angular kinematics and the movement control of the toe vertical trajectory and heel horizontal velocity, strategies to prevent falls by tripping or slipping must address ankle joint kinematics, as also the knee joint kinematics. These alterations on gait kinematics due to fast-walking were followed by changes in muscular activation pattern of the lower limb. Measures of interstride variability of muscle activation for TA and GM reduced, and regions of intra-stride variability changed during fast-walking. These neuromuscular alterations were consistent with those of the angular kinematics. Moreover, the reductions observed in muscular activity, levels of coactivation and variability in muscular activation pattern could be interpreted as an attempt to reduce metabolic consumption during the activity.

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Chapter 9 Conclusions and Suggestions for future research

The current research sought to investigate the effects of fast-walking over time on gait. Kinematic and electromyography data were analyzed to describe the motor and neuromuscular adaptations caused during the activity. Based on the findings obtained in the studies presented in this Thesis, it seems reasonable to stress out the following conclusions:

- Throughout the activity, alterations at the hip and ankle of old adults suggest a redistribution of joint torques and its relative contributions to the total performance;
- Incipient signs of fatigue could be noticed in both groups;
- During the activity, both age-groups showed adaptations linked to strategies of reduction in energy consumption;
- Assessment of interstride variability of angular kinematics showed no effect of the activity over time. However, different values of variability were found in the joint angles, which suggest that the motor output may have more degrees of freedom in one of the components of the kinematic chain;
- Fast-walking induced changes in regions of higher variability within the gait cycle. Revealing that the adaptation of the motor system to a perturbation is phase-dependent;
- Fast walking may increase risk of falls by tripping or slipping;
- Taking all together, our findings revealed that effects of fast-walking on gait were more prominent during the swing phase of the gait. Interestingly, it is during this phase that a fall is more likely to happen.

In summary, during the exercise walking pattern changed over time for both age-groups. Fast-walking induced adaptations consistent with incipient signs of fatigue. Changes in intra-stride variability suggest a shift in the motor strategy during the activity. Although speculative, the reduction in the variability of the toe vertical displacement and the heel horizontal velocity (both task-relevant variables) seemed to be achieved at cost of increased variability of the executive variables. Evaluation of variability within the gait cycle provides further information regarding biomechanical adaptations due to fast-walking. The movement variability of the toe trajectory, and the heel velocity showed to be influenced by the variability at the ankle and the knee. These observations and findings of previous studies suggest that measurements to prevent falls from tripping or slipping should focus on the ankle and knee stability.

Suggestions for future research

The present research contributes to the body of literature by providing additional information regarding the ongoing effects of fast-walking activity on gait. Further studies in these areas will make an important contribution to the established body of research into falls prevention.

Future research should concentrate on:

- analysis regarding the effects of fast-walking on level ground;
- analyzing kinetic alterations induced by this activity;
- increasing sample size to improve the reliability of results in future investigations;
- extensive screening of elderly subjects to ensure homogeneity among the participants;

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